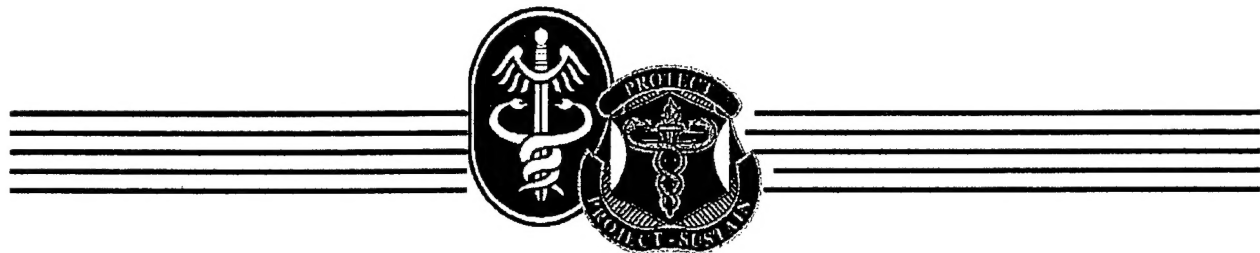


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THE EFFECTS OF WALKING SPEED ON THE BIOMECHANICS
OF BACKPACK LOAD CARRIAGE

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**EFFECTS OF WALKING SPEED ON THE BIOMECHANICS OF BACKPACK LOAD
CARRIAGE**

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BACKGROUND

Only a limited amount of research has been directed toward the study of load carriage, and of that, only a small percentage has been focused on the effects of walking speed. Yet quantitative biomechanical analysis of the effects of walking speed on load carriage can potentially contribute to the effectiveness of equipment evaluation and design. Therefore, the analysis described in this report was undertaken to generate information on the effects of load carriage walking speed on gait kinematics and kinetics in order to form the basis of recommendations concerning pack systems, physical training programs, and load carriage technique. The resulting information could benefit people who engage in load carriage for whatever purpose by increasing load capacity and transport speed, lessening the likelihood of injury, improving efficiency, and decreasing perceived level of difficulty.

In 1984, a biomechanics research program was established at USARIEM. Because of its relevance to the Army, load carriage was selected as a major area of focus of the program. The study described in this report was initiated to increase knowledge about the effects of the weight carried and the speed of walking on the kinematics and kinetics of gait, and on the pattern and degree of muscle involvement revealed by electromyography (EMG). An analysis of the effects of the amount of weight carried on gait biomechanics has already been performed and was reported in a previous USARIEM technical report (25). The present report provides further analysis of the study data to cover the effects of walking speed on the biomechanics of load carriage. In addition, the report provides analysis and discussion of the statistical interaction between weight carried and walking speed.

LIST OF SYMBOLS, ABBREVIATIONS, AND ACRONYMS

mph	miles per hour
USARIEM	U.S. Army Research Institute of Environmental Medicine
USASBCC	U.S. Army Soldier, Biological, and Chemical Command
USASSC	U.S. Army Soldier Systems Center in Natick, MA
ALICE	All purpose, lightweight, individual carrying equipment

EXECUTIVE SUMMARY

To gather information on the effects of backpack weight on gait biomechanics, we analyzed load carriage gait with a cinematographic system, force platform, tri-axial accelerometer, and six surface muscle electrodes. Sixteen male volunteers walked at 1.17, 1.33, and 1.50 m/s, corresponding to slow, medium, and fast speeds, while carrying backpacks ranging from light (6 kg) to very heavy (47 kg) while electrical activity of the trapezius, spinal erector, quadriceps, hamstrings, gastrocnemius, and tibialis anterior muscles were monitored. Based on the data analysis, the following conclusions were drawn: 1) As load carriage speed increased a) there was greater knee flexion at heel-strike, which probably helped reduce shock; b) hip position at toe-off became more extended, as the rear leg pushed off to a greater degree and the front leg stretched further forward; c) there was greater total arm swing, most of which was accounted for by increased arm swing in the rearward direction, which helped prevent excessive trunk rotation; d) the minimum vertical position of the body center of mass declined, reflecting the geometry of a longer stride length; e) there were greater upward and downward center of mass vertical velocities, necessitated by greater stride frequency and vertical center of mass range of motion; and f) changes occurred in load carriage technique that kept several ground reaction forces lower than proportional to the increase in speed. 2) The greatest percentage of joint torque increase with load carriage speed increase occurred about the hip, and the least occurred about the ankle, indicating that muscles producing torque about the hip were most involved in increasing load carriage speed and those producing torque about the ankle were the least involved. The electrical activity data from the leg muscles supported the joint torque findings. Increase in load carriage speed was effected much more through increasing horizontal than vertical ground reaction force. 3) While amplitude of muscle activity tended to increase with speed, the patterns of muscle activity remained the same. 4) The initial propulsive impulse seen at heel-strike resulted from flexion at the knee, rather than extension at the hip, as the heel struck the ground, and was effected by hamstring muscle activity. 5) The pack frame and waist belt did not prevent the shoulders from supporting a considerable portion of the load. 6) The spinal erectors were most active at contralateral heel-strike, when they controlled trunk twisting during the stride. 7) A smaller burst of spinal erector activity at ipsilateral heel-strike was related to deceleration of the forward motion of the trunk as the body was braked at heel-strike, and the subsequent raising of the trunk. 8) The load carriage stride was characterized by a) concentric knee flexion from shortly before to shortly after heel-strike, eccentric knee flexion during a shock absorption phase, concentric knee extension during push-off, and a quiescent period after toe-off during the swing phase; b) eccentric tibialis anterior activity at heel-strike which controlled the rate of plantarflexion to prevent the foot from slapping against the ground, and increased proportionally to speed. The gastrocnemius was largely inactive except for a period of high activity during push-off, which occurred between mid-stance and heel-strike of the contralateral foot and included the second peak for vertical ground reaction force. The report includes recommendations about load carriage technique, physical training exercises to improve load carriage performance, load carriage equipment design, and directions for future study.

DEFINITION OF TERMS

For readers interested in the biomechanics of load carriage, but unfamiliar with its terminology, the definitions below will be helpful for understanding this report:

1. **Stride Time.** The time for a full stride, which includes both a left and a right step. We measure stride time as the time between consecutive right heel-strikes.
2. **Stride Length.** The length of a full stride, which includes both a left and a right step. We measure stride length as the horizontal distance between the locations of two consecutive right heel-strikes.
3. **Stance Phase.** When a given foot is in contact with the ground. It begins with the foot's heel-strike and ends with its toe-off. Each complete gait cycle includes a stance phase for each foot. The stance phase makes up about 60% of the walking gait cycle with little variation for age and height at normal walking speed (40, 49).
4. **Swing Phase.** When a foot is not in contact with the ground. It begins with the foot's toe-off, continues as the foot swings forward, and ends with its heel-strike. Each complete gait cycle includes a swing phase for each foot. The swing phase makes up about 40% of the walking gait cycle, with little variation for age and height at normal walking speed (40, 49).
5. **Single-support.** The period during a gait cycle when only one foot is in contact with the ground; i.e., one foot is in its stance phase while the other foot is in its swing phase. A single-support period of the right foot begins at toe-off of the left foot and ends at the subsequent heel-strike of the left foot. Each complete gait cycle includes a single-support phase on each foot.
6. **Double-support Phase.** The period during a gait cycle when both feet are in contact with the ground at the same time; i.e. both feet are in their respective stance phases. Each complete gait cycle includes two double-support phases. One begins as the right heel strikes the ground, while the left foot is still on the ground. It continues as weight is shifted from the left foot to the right foot and ends when the toe of the left foot leaves the ground. The other begins as the left heel strikes the ground while the right foot is still on the ground. It continues as weight is shifted from the right foot to the left foot and ends when the toe of the right foot leaves the ground.
7. **Ground Reaction Force.** The force exerted by the ground on the foot, which is equal in magnitude and opposite in direction to the force exerted by the foot on the ground.
8. **Joint Torque.** The net impetus exerted by the muscles around a joint to rotate adjacent body segments towards or away from each other around the joint; it is quantified as the muscle force times the perpendicular distance from the line of action of the force to the pivot point of the joint.
9. **Impulse.** The area under the curve of force as a function of time.
10. **Kinematics.** Quantification of motion without regard for the forces producing the motion. Human kinematic data include linear and rotational position, velocity, acceleration, and range of motion for each body segment and the total body center of mass. It also includes such variables as stride length, stride frequency, and relative time in single and double-support.

11. **Kinetics.** Analysis of the forces and torques that bring about motion. Human kinetic data include ground reaction forces, joint bone-on-bone forces, and muscle torques.
12. **Electromyography.** Recording and analysis of muscle electrical activity.

INTRODUCTION

KINEMATIC ASPECTS OF GAIT

Studies of human gait are inherently complex because of the interrelationships among the various parameters that describe walking and running. For example, several studies have shown that many gait parameters are velocity-dependent. Grieve (22) reported on the relationship between time-distance parameters and walking speed. Other investigators (14, 39) have reported that angular limb motion, muscular activity, and joint reactions are dependent on walking speed.

Effects of Walking Speed

The adaptation to increasing speed is accomplished by increases in both the frequency and amplitude of leg movement (22, 23). As the speed of gait increases, so does the range of motion of lower extremity joints (26, 35). The percentage of time for the stance phase progressively decreases from 62% for walking to 31% for jogging and 22% for sprinting (35). The center of gravity of the body becomes lower as locomotion speed increases due to increased flexion of the hips and knees and greater dorsiflexion at the ankle joint (35).

Murray (39) reported that adult males decreased their stance time 3.5 times as rapidly as their swing time when cadence increased. Grieve and Gear (22) developed regression equations relating swing time to velocity and stride length. Murray et al. (39) compared walking patterns of 30 healthy men with normal range of motion who were asked to walk at both a comfortable pace and a fast pace. The results of the study showed that average walking speed was 2.12 m/s for fast walking and 1.51 m/s for comfortably paced walking. For fast and free walking respectively, stride time was 0.87 s and 1.06 s; stride length was 1.86 m and 1.56 m; percentage of gait cycle in the stance phase was 57% and 61%; and percentage of gait cycle in double-support was 14% and 22%. Thus, as walking pace quickens, stride time shortens, stride length increases, the percentage of the stride in stance increases, and the percentage of the stride in double-support increases.

According to Winter (58), joint angle patterns over the stride period were quite consistent, regardless of stepping cadence. He studied the hip, knee, and ankle joint on three groups of volunteers, one of which was instructed to walk at a natural cadence (N=16), another of which walked at a slow cadence (20 steps/min less than their natural cadence, N=14), and a third of which walked at a fast cadence (20 steps/min more than their natural cadence, N=14). The results showed consistent joint angle patterns that were relatively insensitive to walking cadence. Toe-off was seen to occur at about 62% of stride time. There were minor, speed-related differences immediately after heel contact: the slow cadence walkers plantarflexed to 6° from the neutral position (foot at 90° from the shank), the natural cadence walkers to 3.5°, and the fast cadence walkers to 1.5°. Knee angle showed more variation. The slow cadence walkers flexed to 14°, the natural cadence walkers to 20°, and the fast cadence walkers to 24° early in the

stance phase. The knee reached its maximum degree of flexion (62° - 67°) at 73% of stride. The hip angle patterns were almost identical at all cadences.

Several studies have been focussed on the effects of changes in walking speed on cadence and stride length (16, 39, 40, 53). Murray et al. (40), in a study on 60 normal males, found a mean walking cadence of 117 steps per minute, which is faster than the rate observed by Winter (59). Mean stride length was 156.5 cm for a volunteer population ranging in age from 25 to 65 years. The authors concluded that age, height, trial number, and side (right or left) had no systematic influence on cadence. However, cadence decreased slightly after the age of 40, and stride length decreased after 60 years of age. Some researchers (4, 27) have reported that the natural walking cadence is the most efficient one.

KINETIC ASPECTS OF GAIT

Ground Reaction Forces

To kinetically analyze performances in which two parts of the body come into contact with an external object (which may include the ground), it is necessary to directly measure the force exerted by at least one of those body parts on the external object. This applies to activities such as walking, manual labor, and load carriage. The necessary information cannot be inferred from movement studies alone, using methodology such as goniometry or cinematography.

During running, no more than one foot makes contact with the ground at any given time. Thus it is possible to calculate forces and torques on the body from kinematic data and knowledge of the volunteer's body mass. However, during walking, both feet contact the ground at the same time during the two double-support phases of each full stride. Therefore, force platform data are required to enable a kinetic analysis. As the foot exerts force on the ground during the stance phase of a stride, the ground exerts equal and opposite force on the foot. The study of ground reaction forces during walking can provide relevant information about the mechanics of gait under various conditions. It provides a direct measure of impact forces on the foot, and thus is relevant to the understanding and prevention of lower extremity injuries.

Force platforms, which use sensing elements whose electrical characteristics change in proportion to the magnitude of applied forces, are used to measure the forces and moments applied by the foot on the ground. If a complete force and torque record of a footstep is to be obtained, each of the force and moment components must be sampled at a sufficiently high rate. An example of the use of force platform technology is the diagnosis of hip joint problems through evaluation of the vertical component of ground reaction force during walking, decomposition of the force into the sum of its harmonic components, and description of the force in other mathematical terms (28). Bresler and Frankel (11) studied different characteristics of vertical ground reaction force measured on a force platform. Yamashita and Katoh (61) used a

specially designed force platform to analyze the pattern of center of pressure during level walking.

Schneider and Chao (46) analyzed the ground reaction forces of 26 normal volunteers during walking. The curve of vertical ground reaction force as a function of time typically had a dual-hump shape with the second peak higher (114% of body weight) than the first (106% of body weight). When graphed as a function of time, vertical ground reaction force formed a pattern that was nearly symmetrical about a vertical line at 50% of the stance phase of each foot. The front-back ground reaction force was not symmetrically distributed, with a larger peak forward (propulsive) force (19.0% of body weight) and a smaller peak rearward (braking) force (15% of body weight). The waveform of the medio-lateral ground reaction force was more irregular than that of the other two ground reaction forces. The medial ground reaction force was predominant except at both ends of the stance phase.

Walking speed affects ground reaction forces. Jansen and Jansen (29) observed a general increase in vertical ground reaction forces at both heel-strike and toe-off with increasing speed, and a significant decrease in force during the mid-stance period. The anterior-posterior forces were also observed to increase proportionally with increasing speed, but there was no significant change in medio-lateral reaction forces.

Soames and Richardson (50) evaluated the influence of stride length and cadence on ground reaction forces during gait. Recordings of the three ground reaction forces were taken from 12 volunteers (six males and six females), aged 19 to 23. Stride length was standardized as 50%, 75%, or 100% of the volunteer's leg length. The step frequency was constrained to 42, 52, or 62 steps/min. Stride length and cadence significantly influenced all three ground reaction forces at heel-strike, while cadence influenced the anterior-posterior ground reaction force at toe-off. The effects of increasing cadence were more pronounced than those of increasing stride length.

Andriacchi, Ogle, and Galante (3) observed gait of normal volunteers and patients with knee disabilities. They reported time-distance measurements and ground reaction force parameters in relation to walking speed (slow, normal, and fast walking). They used regression analysis to establish functional relations between ground reaction force amplitudes and walking speed. The results showed that all ground reaction forces were affected by walking speed and that the minimum vertical ground reaction force was more affected by change in walking speed than was any other ground reaction force variable.

Joint Moments And Forces

An understanding of the effects of forces on material bodies is essential to the study of locomotion. The strength of a rotational impetus is called moment of force and is equal to the magnitude of the force multiplied by the perpendicular distance from the line of action of the force to the point of rotation. A kinetic analysis of walking (2, 46)

and running (36, 38, 57) revealed basic patterns of moments generated by the muscles around the ankle, knee, and hip. However, individual differences in pattern of moments about the knee and hip during gait have also been noted (43, 57).

Simon et al. (48) investigated the forces generated at heel-strike during human gait using both a force platform and a force transducer inserted into the heel of the shoe. The output traces were analyzed for the existence of high frequency impulsive loads during a normal walking cycle. The data showed that during normal human gait the lower limb is subjected to a high impulsive load at heel-strike. The severity of this impulse varied with the individual, the walking velocity, the angle with which the limb approached the ground, and the compliance of the two materials coming in contact at heel-strike. Peak force varied from 0.5 to 1.25 times body weight, and its frequency components varied from 10 to 75 Hz.

Paul (42) has shown that average joint forces of both the hip and knee joints increase with increasing stride length, while Stauffer, Chao, and Brewster (54) observed that an increase in cadence during free walking in normal volunteers did not significantly change the magnitude of the peak compressive forces across the ankle joint.

Several other studies (36, 38, 57) revealed basic moment patterns during running. Winter (57) studied ankle, knee, and hip moments while 11 normal volunteers jogging at slow speed. He found that the moment of force for the total lower limb was primarily extensor during the stance phase. He also noted the relative timing of the peak extensor torques at the three joints. The hip peaked at 20% of stance, the knee at 40% of stance, and the ankle near 60% of stance. The variability of the moment patterns across all jogging trials was considerably less than that seen during walking. Two power bursts were seen at the ankle, including an absorption phase early in the stance followed by a dominant generation peak during late push-off. Average peak power generation was 800 W with individual maximums exceeding 1500 W.

ELECTROMYOGRAPHIC ANALYSIS OF GAIT

Over the years electromyography (EMG) has been used to investigate the activity of the muscles of the lower extremity during walking. It provides a recording of muscle electrical activity between two conducting electrodes, which vary in type and construction. The two main types of EMG electrodes are surface electrodes and indwelling (needle and wire) electrodes. Each has its advantages and its disadvantages.

The needle electrode is the most common type of indwelling electrode used for clinical diagnostic purposes but is unsuited for studies of movement. One of two main advantages of the needle electrode is that its small pickup area enables the electrode to detect individual motor unit action potentials during low-force contractions. The other advantage is that it may be repositioned within the muscle so that new territories may be explored (7). It can pick up EMG signals from muscle fibers up to 1.5 mm away. The

needle electrode can be used to detect signals from deep muscles, and receive signals from a much more confined area than surface electrodes. In order to obtain EMG signals from an entire muscle group, a number of electrodes would have to be used, which would reduce the volunteer's comfort.

The fine wire electrode is another type of indwelling electrode. It is extremely fine and easily implanted and withdrawn, therefore painless. The main purpose of using wire electrodes in human movement studies is to record a signal that is proportional to the contraction force of a muscle. A limitation of the wire electrode is its tendency to migrate after it has been inserted during the first few contractions of the muscle. Basmajian and De Luca (7) suggested that muscle with the electrode be contracted and relaxed at least one-half dozen times before any measurements are taken.

Surface electrodes may be used effectively with superficial muscles but, because they pick up signals from a broad area of muscle near the skin, cannot be used to detect signals from small, deep muscles. The main advantage of surface electrodes is that they are convenient to use and provide high fidelity EMG signals. Surface electrodes are acceptable when the time of activation, frequency, and magnitude of EMG signals are to be examined, but small and/or deep muscles are not the objects of interest.

Some studies have been undertaken which used electromyography to examine neural control of gait (10, 18, 43). The results of those studies provided some indication of when certain muscles are on and off during the gait cycle but have not given quantitative measures of the intensity of muscle activation. There are significant changes in EMG timing and magnitude as walking speed changes (31).

During level walking, the hamstrings and tibialis anterior reach peak activity at heel-strike. Quadriceps muscle activity increases thereafter to keep the knee from buckling and then to push off. The hamstrings and quadriceps show elevated EMG activity starting just before and continuing until just after toe-off (33). The calf muscles increase their activity gradually from the mid-stance phase until toe-off. The knee stabilizing function of the gastrocnemius is most important during the stance phase (55). The calf muscles are active during knee extension and ankle dorsiflexion during the mid-stance phase (20). Even after quadriceps activity ceases during the mid-stance phase, knee extension continues, due to torque about the knee resulting from movement of the upper body center of mass forward of the knee joint (40).

Effects of Walking Speed on Muscle Electrical Activity

EMG activity of the quadriceps, hamstrings, tibialis anterior, and gastrocnemius increases significantly with walking speed. Winter (58) evaluated EMG patterns during treadmill walking of eight normal volunteers at natural (2.25 mph = 1.0 m/sec), slow (1.75 mph = 0.78 m/sec), and very slow (1.25 mph = 0.56 m/sec) walking speeds. Five muscles were assessed: soleus, tibialis anterior, rectus femoris, vastus lateralis, and hamstrings. For all muscle groups, the patterns of activity remained essentially the

same at all walking speeds, but EMG amplitude increased with walking speed. Soleus and tibialis anterior muscle activity dropped about 30% as the volunteers slowed down. Greater reductions in activity were evident by the muscles crossing the knee and hip: vastus lateralis activity decreased by about 50%, while rectus femoris and hamstring activity dropped by about 70%.

Mann (35) performed a biomechanical study of 13 runners including two male sprinters, five experienced joggers, and six elite long distance runners. The muscles monitored about knee joint showed no difference in the function of the various quadriceps muscles during the stance phase. During walking, the quadriceps became active at approximately the last 10% of the swing phase and remained active during the first 15% of the stance phase. During running, the quadriceps became active in the last 20% of the swing phase and remained active for the first 50% of stance phase. Posterior calf muscles during walking functioned approximately during the middle 50% of stance phase. During running this period of activity increased considerably. Anterior compartment muscles became active during walking before toe-off and remained active during the entire swing phase and through the first 10% of the stance phase. During running, activity began at the time of push-off and continued through the first 80% of the stance phase.

PHYSIOLOGICAL CORRELATES OF GAIT MECHANICS

The physiological responses of volunteers carrying loads, especially as to energy cost, have been examined in some detail. Energy cost increases in a systematic manner with increases in body weight (19, 21), load (6, 9, 17, 32, 51, 52), velocity (52) and grade (9, 21, 41).

Mechanical analysis of walking has been studied for several years (15). Cavagna and Margaria (13) introduced energy calculations from force platform data, with the body regarded as a point mass. Winter, Quanbury, and Reimer (60) developed a mechanical energy calculation method based on a segment-by-segment analysis assuming energy exchanges within segments and energy transfer between adjacent segments.

PURPOSE OF THE PRESENT STUDY

This study was undertaken to increase knowledge about the biomechanics of load carriage. While previous studies of load carriage produced important information, many questions remained unanswered. This study provided new information by looking at the effects of three prescribed walking speeds, from leisurely to fast. It also enabled determination of whether the effects of walking speed were consistent across different backpack loads in that the volunteers carried four different loads at each walking speed. The use of both cinematographic analysis and electromyography provided the opportunity to calculate both body kinematics during load carriage and the joint torques generated by muscle groups needed to effect the observed body movements. This

allowed assessment of which muscles were active and their degree of involvement. The military relevance of the study was enhanced by the fact that most of the volunteers were U.S. soldiers and the backpack frame, waist belt, and shoulder straps were from the standard Army ALICE backpack, which had been used by U.S. soldiers for decades.

METHODS

VOLUNTEERS

Testing occurred at the biomechanics laboratory of the U.S. Army Research Institute of Environmental Medicine, Natick, MA. Volunteers for the experiment included permanent party military test volunteers assigned to the U.S. Army Natick Soldier Center, soldiers recruited for temporary duty as test volunteers, and military and civilian employees of the U.S. Army Research Institute of Environmental Medicine. A total of 16 volunteers were tested. Each volunteer participated in the study for a maximum of two test conditions per day.

Sample Size Estimation

A nomogram for repeated measures (12) was used to estimate the sample size. The nomogram (Appendix A) shows the minimum difference between the dependent variable means of the experimental groups, in standard deviation units, that can be found statistically significant, given a Type I error rate of 5%. The nomogram has two vertical scales, sample size on the left side and inter-trial correlation coefficients on the right side, with a diagonal scale between them representing the minimum detectable mean difference (effect size). To find the number of volunteers needed, a line is drawn from the inter-trial correlation coefficient through the desired effect size to the sample size scale. For a given effect size, the higher the inter-trial correlation coefficient, the fewer test volunteers were needed. A higher inter-trial correlation coefficient enables the researcher to detect a smaller mean difference with the same sample size.

Because the inter-trial correlation coefficients of most dependent variables analyzed in the biomechanical study of load carriage were available from pilot study, sample size estimation could be performed easily. For example, an inter-trial correlation coefficient of about 0.90 for stride length and stride frequency with effect size of 0.5 gives sample size of smaller than 5. An inter-trial correlation coefficient of about 0.70 for the EMGs with effect size of 0.5 gives a sample size of 10. The inter-trial correlation coefficients for most of the variables examined were higher than 0.60. The nomogram showed that for an inter-trial correlation coefficient of 0.60 with a moderate effect size of 0.5 and a two-tailed alpha level of 0.05, 13 volunteers were needed. It was decided to test 16 volunteers in order to provide for data lost by equipment malfunction or volunteers who might terminate testing prematurely.

INSTRUMENTATION

Force Platform System

Information needed for the kinetic analysis of load carriage includes the forces exerted by the ground on the feet (ground reaction forces). A force platform provides the needed information because the ground reaction forces are equal in magnitude to, and opposite in direction from, the forces exerted by the feet on the force platform.

Information provided by the force platform includes the magnitudes of forces exerted by the feet in the vertical, front-back, and left-right directions relative to the walker as well as the location on the platform of the foot center of pressure. Knowledge of the latter is essential in order to calculate the moment about the ankle joint due to ground reaction force, which is directly proportional to the distance from the point of application of the force to the joint. In addition, error in calculation of torque about the ankle results in errors in torque calculations for the knee and hip, since calculations are performed in sequence from the ankle up.

A model LG6-1-1 force platform from Advanced Mechanical Technology Incorporated (Newton, MA) was used in conjunction with a model SGA6-3 amplifier designed for use with computerized data acquisition systems. The plate, which measures .61 by 1.22 m (two by four feet), was mounted on a steel frame to keep it rigid and isolated from external vibrations that might cause spurious output signals. The no-damage limits of the platform were 2,200 pounds (9,800 N) of vertical load applied anywhere on the top surface or 1,200 pounds (6,700 N) of horizontal load applied perpendicular to any of the platform's sides. The system was designed to emit voltage signals proportional to forces and torques exerted on the plate's surface, which include forces in the vertical, front-back and left-right directions and torques around orthogonal axes through the center of the plate oriented in the latter three directions. Center of pressure can be calculated from the forces and torques, as specified in the AMTI force platform manual. The force platform and walking surfaces were made flush by building a wooden platform around the force platform. The SGA6-3 amplifier system contained a six-channel amplifier with switch-selectable gains of 1000, 2000, and 4000 for each channel. Each channel also had a selectable low-pass filter with a 10 Hz or 1,050 Hz cutoff frequency and selectable precision bridge excitation voltages of 2.5, 5, or 10.

Accelerometer

A model EGAXT3-84-c-100 tri-axial accelerometer (Entran Devices, Fairfield, NJ) was mounted in the pack during load carriage. It emitted voltage signals proportional to pack acceleration in three orthogonal directions. This temperature compensated strain gauge accelerometer measured accelerations in the range of ± 100 g in the vertical, left-right, and front-back directions. Built-in over-ranging protection prevented damage to the device. Because of a very high resonant frequency of 1,700 Hz, the accelerometer did not distort the accelerations characteristics of human movement.

Cinematography System

Cinematography has been used for several years for the biomechanical analyses of gait (8, 30, 37, 44). The process involves filming human movement with one or more cameras driven by spring, battery, or line power, at a frame rate fast enough to capture the movement with adequate resolution. In contrast to video systems, 16 mm film requires a considerable amount of light, especially indoors and at high frame rates. After the film was processed, it was projected frame by frame onto a digitizing table.

where the experimenter used a pointing device to locate major joint centers of the body. The digitizing device sent table coordinates of the joint locations to a computer with which it was interfaced. Computer programs then processed the coordinate information to calculate kinematic variables that included body segment positions, velocities, and accelerations. The volunteer's body mass and data from a force platform were processed along with the kinematic data to produce kinetic information, which included the forces and torques at each body joint.

Video analysis has supplanted cinematography to a large extent, mainly because digitizing can be accomplished automatically, eliminating the slow and tedious process of hand-digitizing film images. Video also has the advantages of immediate availability of collected data and the low cost and reusability of videotape. However, current video systems cannot rival the image resolution of 16 mm film.

One LOCAM II camera from Redlake Corp. (Morgan Hill, CA) was used to film the volunteers during load carriage. The camera can be set at precise frame rates up to 500/sec. A frame rate of 60 Hz was used for this experiment because it was fast enough to capture the body movements of interest. A faster frame rate would unnecessarily require more film and more time spent in film digitizing. The camera incorporates a timing light which places markers on the edge of the film every .01 sec to allow checking of film speed. A model 12-0101 battery pack permitted use of the camera away from AC power outlets. Model 9003-0001 floodlights (1000 watts) from Colortran (Burbank, CA) and model 18001 Mini-Mac photoflood lamps (1000 watts) from Bardwell & McAlister (Hollywood, CA) provided illumination.

For analysis, developed films were projected with an M-16C projection head from Vanguard Instrument Corp., Melville, NY, onto an ACT23 digitizing table from Atek Corporation (Silver Spring, MD). The projector allowed one frame of the film to be seen at a time. Specific frames could be referenced using a digital frame counter. The digitizing table had a resolution of .01 mm and was connected via its controller to a model 486-33 IBM-PC compatible computer from Club American Technology Inc. of Fremont, CA.

Electromyography System

"Utah" model surface electrodes with integral preamplifiers and band pass filtering systems from Motion Control Inc. (Salt Lake City, UT) were used to record muscle potentials from the shoulder, back and legs. Each electrode was factory calibrated, with individual gains ranging from 340 to 380. Although the gain was slightly affected by the frequency of the signal being amplified, the variation in gain for signals between 60 and 500 Hz was within 2% of the range. The bandwidth of the preamplifier was 8 Hz to 33 KHz. The high input impedance of the electrodes made it unnecessary to abrade the skin or use electro-conductive jelly.

Computerized Data Collection System

Data was sent to a model 486-33 IBM-PC compatible computer from Club American Technology Inc. (Fremont, CA), including six output signals from the force platform, three from the accelerometer, six from the muscle EMG electrodes, and one from the event marker, for a total of 15. The signals were fed into a model DAP1200/2 data acquisition and analog-to-digital converter board from Microstar Laboratories Inc. (Redmond, WA) mounted in an expansion slot in the computer. The DAP combined analog data acquisition hardware with a 16-bit microprocessor and a real-time multitasking operating system. It had 16 channels, each of which could be specified in software as single-ended or differential.

The inputs to the DAP were voltages, which the board converted to numbers. The board could perform computations on the resulting numbers before the information was sent to the computer, making data processing very fast. The gain factor was independently software selectable for each channel, with possible values of 1, 10, 100 and 1,000. Allowable voltage input ranges with unity gain were 0 to 5.0 V, -2.5 to +2.5 V, -5 to +5 V, and -10 to +10 V. Maximum sampling rate was 50,000 per second. The sampling rate for this experiment was 1,000 Hz for all the channels except for the EMGs. Two logical channels operating at 1,000 Hz each were used for each EMG hardware channel, so that the actual sampling rate was 2,000 Hz per EMG channel.

Backpack

A backpack (Figure 1) was specially designed for the experiment, using a standard U.S. Army ALICE external pack frame as a base. Two metal shelves were added to the frame. On the bottom shelf was mounted a metal box containing the accelerometer, a terminal for the EMG electrodes, and a junction for a multi-conductor cable through which output data could be sent to the analog-to-digital converter board mounted in the computer.

The top shelf of the pack was designed to hold weights so that the intended experimental loads could be carried in the pack. The weights were in the form of lead bricks and rectangular iron plates.

An effort was made to match as closely as possible the location of the center of mass of the experimental pack and an ordinary backpack. A pack loaded in standard fashion was balanced on a straight edge to locate its center of mass. The weights were then arranged on the experimental pack in such a manner as to match the center of mass location of the standard pack. Blocks of stiff foam were used as spacers on the shelf under the weights to make sure all of the pack loads had the same center of mass.

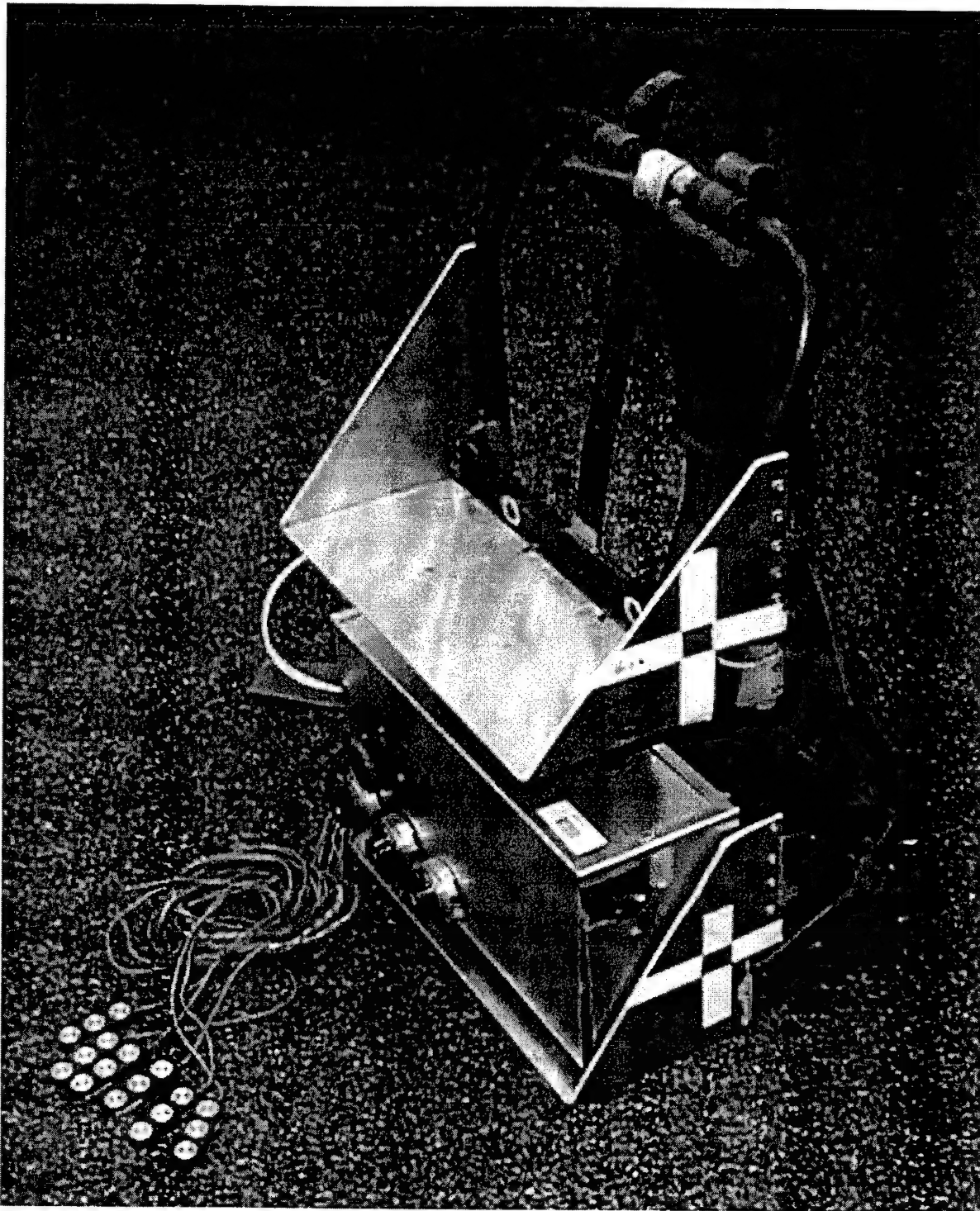


Figure 1. The instrumented backpack used in the experiment.

Two tape markers were placed on the side of the experimental pack so that the pack's position could be determined throughout a filmed trial by digitizing. The location of the actual pack center of mass relative to the markers was measured and recorded for use by the film analysis computer program.

Speed Cuing Device

A device to pace the volunteer's walking speed was designed at the U.S. Army Research Institute of Environmental Medicine and fabricated at the U.S. Army Soldier Systems Center in Natick, MA. It was based on a motor-driven cord marked with alternating light and dark bands that traveled around two pulley-wheels spaced 8 m apart. The speed of the cord was set using a dial. A digital speed display enabled cord speed to be set to the nearest 0.01 m/s. During an experimental trial, the device was oriented alongside the volunteer so that the visible part of the cord traveled in the direction the volunteer walked. The volunteer walked straight ahead while maintaining a peripheral view of the moving cord, which cued the appropriate walking speed. The actual walking speed was determined later from the film data.

EXPERIMENTAL PROCEDURES

Independent Variable

Locomotion Speed. Just as speed affects the technique of walking and running (45), it is likely to affect the biomechanics of load carriage. Locomotion speeds during testing were visually cued at 1.1, 1.3, and 1.5 m/s by the specially designed device placed alongside the volunteer. These can be characterized as slow, medium, and fast marching speeds, respectively.

Dependent Variables

The following variables were calculated from the vertical, front-back and left-right forces exerted by the feet on the force platform:

- a. heel-strike and push-off peak forces (N)
- b. time of occurrence of heel-strike and push-off peak force (% of stride time)
- c. peak and average front-back and mediolateral forces (N)
- d. positive and negative vertical, front-back and mediolateral impulse per stride (N·sec)

Film analysis allowed calculation of the following:

- a. joint ranges of motion for the hip, knee, and ankle (radians)
- b. joint torques for the hip, knee, and ankle (N·m)
- c. joint forces at the hip, knee, and ankle (N)
- d. stride length (m)
- e. stride frequency (strides/min)

- f. single-support time (% of stride time)
- g. double-support time (% of stride time)
- h. body segment and center of mass position, velocity and acceleration

EMG analysis allowed calculation of the following:

- a. peak and average muscle activities for the trapezius, spinal erector, quadriceps, hamstrings, gastrocnemius, and tibialis anterior muscles (uV)
- b. timing of activation for the muscles listed above

Accelerometer data analysis allowed calculation of the following:

- a. peak accelerations of the backpack in the vertical, front-back, and left-right directions (g)
- b. timing and directions of the accelerations

Test Trials

All volunteers were orally briefed on the purpose, risks, and benefits of the study, after which they signed informed consent documents. Electrodes were attached to the volunteers' skin with adhesive tape after the skin was cleaned but not abraded with rubbing alcohol and a gauze pad. Electrodes were placed over the following muscles using anatomical landmarks according to the recommendations for standardized electrode positions (62 and Inter-Med Industries):

- trapezius (elevates the shoulders, resists shoulder depression under the weight of the backpack)
- lower erector spinae, L4/L5 level (extends the back, resists forward movement of the trunk due to backpack weight and inertia)
- rectus femoris (extends the knee and flexes the hip during locomotion, helps lift the weight of body and backpack during the stride)
- biceps femoris (flexes the knee, extends the hip)
- tibialis anterior (works eccentrically to control the speed of foot plantarflexion so that the foot does not slap the ground too quickly)
- gastrocnemius (plantarflexes the foot, helps lift the weight of body and backpack during the stride)

The volunteers performed their test trials (Figure 2) while wearing shorts and military boots. Prior to data collection, reflective tape markers were placed on the right side-view joint centers of the ball of the foot, ankle, knee, hip, shoulder, elbow, and wrist. Volunteers then donned the loaded backpack. Trials consisted of walks of no more than 15 m across the force platform in the camera field of view. Each volunteer was given practice trials to adjust walking speed and starting position so that the right foot landed squarely on the force platform as the volunteer walked across it. Data for the EMGs, force platform, and accelerometer were collected for every trial, but only the data from acceptable trials were saved. A volunteer performed no more than nine trials

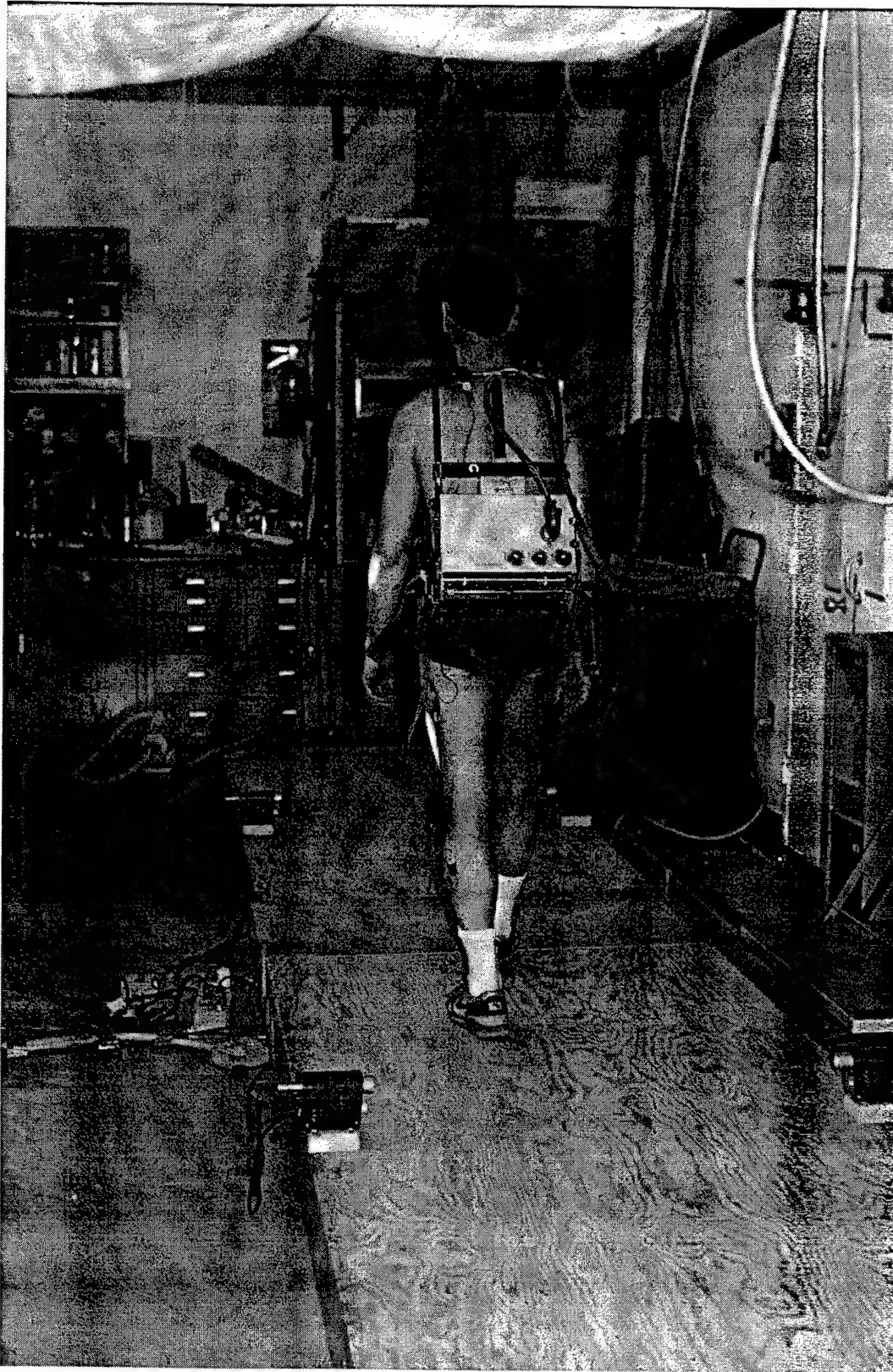


Figure 2. The experimental setup. For the actual trials, volunteers wore military boots.

in a test session (one load x three speeds x three trials), with a maximum of two test sessions per volunteer per day (one in the morning and one in the afternoon). A volunteer carried a different load for each test session resulting in a total of 36 acceptable trials over four test sessions. Occasionally, a trial had to be repeated if the volunteer's foot did not land directly on the force platform. Adequate rest periods were allowed between trials to avoid fatigue as a confounding factor. Each trial lasted no more than 15 seconds, so total exercise time per day was minimal.

Data Processing

Data were collected and analyzed on the computer. A program specifically written for the study in the C++ computer language was used to collect the digitizing table coordinates from each frame of film, as well as the data from the six force platform channels, the three accelerometer channels, and the six EMG electrodes, all converted from analog signals to numerical information by the A/D board. Other programs performed the processing necessary to compute records of dependent variable values over the stride. A large statistical file then was created which contained key variables describing the gait patterns of all the volunteers.

The EMG data underwent digital-to-RMS conversion (47) and other interpretive procedures. The vertical and horizontal forces determined from the force platform divided by the weight of body-plus-load gave vertical, mediolateral and front-back accelerations of the system center of mass. Mathematical integration of the accelerations yielded velocities.

Digitizing. The film sequence selected was of the load carriage trial closest to the target walking speed for the particular combination of load and walking speed. An experimenter obtained the x-y image coordinates of each marker on a volunteer's body over a full stride by a process called digitizing. That process involved projecting the film one frame at a time on the rear side of the translucent digitizing table. The experimenter sequentially placed the cross-hairs of a transparent mouse-like device over the center of each joint marker image. When the experimenter pressed a button on the device, the x-y digitizer table coordinates of the marker were sent to the computer. A custom-written Borland C++ computer program collected film data from the digitizing table via an IEEE-488 interface board (Capital Equipment Corp., Burlington, MA) installed in one of the computer's expansion slots. The program drew a stick figure of the volunteer on the computer screen as the film was digitized to allow immediate detection and correction of gross digitizing errors. The computer displayed the name of each joint as it was to be digitized. If a digitizing error was made, the program allowed the user to go back and re-digitize any point at will. The program allowed a user to stop digitizing at any time, shut down the computer, and resume again at any time.

The ball of the foot, ankle, knee, hip, shoulder, elbow, wrist, and ear lobe of the right side of the volunteer were digitized. The first frame digitized was 11 frames before the frame at which the right heel passed the back of the left lower leg. The last frame digitized was 12 frames after the right heel again passed the back of the left lower leg.

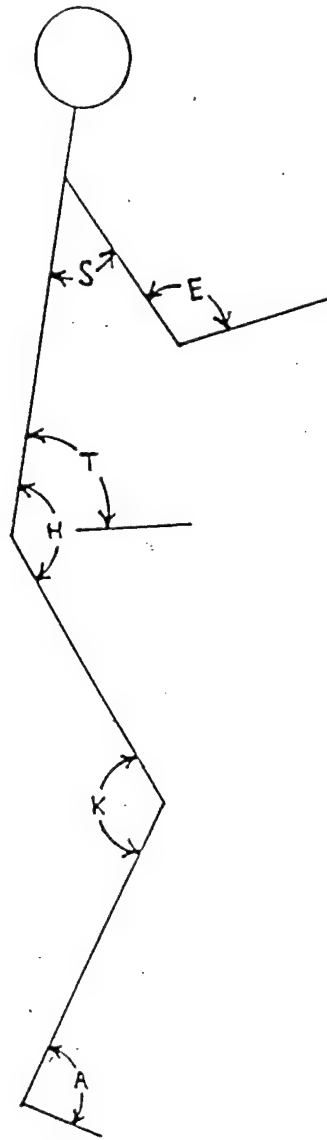
Because the camera was aimed across the center of the force platform, this provided the best film images of a full stride. The extra frames digitized at the beginning and end of the stride were needed for mathematical data smoothing and to ensure that a full stride was recorded. At the beginning of processing the film data from each trial, the four corners of the force platform were digitized in order to later be able to calculate the film coordinates of the center of pressure.

Data Smoothing and Interpolation. The digitized film data were smoothed using Fourier Analysis and Digital Filtering subroutines contained in Software for Science and Engineering Tools IPC-TC-006 (Quinn-Curtis, Needham, MA). The smoothed data were then processed with a cubic spline curve-fitting subroutine from the same software library to produce 101 interpolated frames for one full stride representing 0% to 100% of the time of a full stride. Thus, the results for each volunteer were in terms of percentage of stride. The actual time between interpolated frames was unique to each trial and was later used to calculate actual velocities and accelerations of the body segments and center of mass.

The mass, center of mass, and moment of inertia of each body segment were estimated using tables of standard body proportions based on dissection of cadavers (56). Because both heel-strike and toe-off were visible in the films and on the display of force-platform data, these two points were used to time-synchronize film and force-platform data. The EMG and accelerometer data were already time-synchronized with the force-platform data because they all were concurrently digitized by the computer's analog-to-digital converter board. The foot's center of pressure location on the force-platform's surface was calculated for each trial from force platform data using equations provided by the force-platform's manufacturer (1). Joint moments and forces for the lower extremity were calculated using segment-by-segment kinetic analysis (56).

System of Postural Analysis. To analyze posture throughout the stride, the system of body angles shown in Figure 3 was used.

Statistical Analysis. The large statistical file containing the key variables describing the gait patterns of all the volunteers was transferred to a VAX 780 mainframe computer where programs from BMDP (Berkeley, CA) were used for statistical comparisons between the different experimental conditions. An analysis of variance with repeated measures was performed on each of the variables using the BMDP 2V program. Means and standard deviations for each variable under each testing condition were calculated. A three by four factorial analysis with three levels of speed and four levels of load was performed. Significance of speed/load statistical interaction was tested to see how the effect of load might differ with speed or the effect of speed might differ with the load. Post-Hoc Tukey tests were employed to locate the differences between treatment means when significant treatment effects were found by analysis of variance.



- A = Ankle angle: the absolute ventral angle between foot and shank. Because the foot segment endpoints were the lateral malleolus and ball of the foot, the ankle angle with the bottom surface of the foot at 90° to the shank was about 120°.
- K = Knee angle: the absolute dorsal angle between shank and thigh.
- H = Hip angle: the absolute ventral angle between thigh and trunk.
- T = Trunk angle: the ventral angle between the trunk and a horizontal line.
- E = Elbow angle: the absolute ventral angle between upper arm and forearm.
- S = Shoulder angle: the angle between upper arm and trunk (positive with the upper arm in front of the trunk and negative with the upper arm behind the trunk).

Figure 3. The system of body angles used to analyze posture throughout the stride.

RESULTS

TEST VOLUNTEER CHARACTERISTICS

The test volunteers were all physically fit males, a bit above average in both height and body mass (Table 1). All engaged in regular physical activity. Of the 16 volunteers, 11 were enlisted U.S. Army personnel, 3 were Army officers, and 2 were civilian employees of the U.S. Army Research Institute of Environmental Medicine.

Table 1. Physical characteristics of the test volunteers (means \pm SD)

Age (yr)	30.3 \pm 9.2
Height (cm)	181.2 \pm 7.5
Body mass (kg)	76.8 \pm 8.9
Gender	all male
n	16

SPEED EFFECTS

Actual Walking Speeds

Because there were three trials at each combination of load and speed, preliminary film analysis was used to select the trial at each condition that came closest to the nominal speed. It was determined that the volunteers deviated somewhat from the visually cued walking speeds of 1.1, 1.3 and 1.5 meters per second. They apparently had difficulty keeping their walking speed down to the slowest experimental pace of 1.1 m/s. Of the set of trials selected for final analysis, it was found that the volunteers cued to walk at the slowest speed of 1.1 m/s actually walked at 1.17 ± 0.06 m/s. The volunteers cued to walk at the medium speed of 1.3 m/s walked only slightly faster than the cued pace, at 1.33 ± 0.05 m/s. The volunteers cued to walk at the fast speed of 1.5 m/s were right on target, actually walking at 1.50 ± 0.06 m/s.

Because the actual walking speeds deviated from the cued speeds, the means of the three walking speeds differed by about 0.17 m/s instead of the planned 0.20 m/s. Thus, the increases in speed from slow to medium to fast were in steps of about 13%-14% instead of the planned 15%-18%. Even though the walking speeds were not exactly as planned, they still corresponded to slow, medium, and fast load carriage speeds and likely represented a natural range of speeds for soldiers carrying loads.

Stride Parameters

Table 2 displays means of various stride parameters for the three load carriage speeds; included are results of repeated measures analyses of variance and post-hoc Tukey tests for Honestly Significant Differences (HSD). There were significant speed effects for stride time, stride length, stride frequency, percentage of stride at toe-off, and percentage double-support. Stride time, percentage double-support, and percentage of stride at toe-off decreased significantly as speed increased, with significant differences among all speeds. Stride length and stride frequency increased significantly as speed increased, with significant differences among all speeds.

Stride time decreased as load carriage speed increased, with both stride length and frequency increasing linearly and accounting almost equally for increased walking speed. Stride length and frequency both increased by about 7% as speed increased 14% from 1.17 to 1.33 m/s; stride length and frequency increased another 6% as speed increased 13% from 1.33 to 1.50 m/s. The percentage of stride at which toe-off occurred, which is equivalent to the percentage of stride that each foot is in contact with the ground (stance phase) decreased from 65.2% to 63.6% as test speed increased from 1.17 to 1.50 m/s. Percentage of stride under double-support, which is directly related to the percentage of stride that each foot is in contact with the ground, decreased significantly as well.

Table 2. Stride parameters (mean \pm SD) at load carriage speeds of 1.17, 1.33, and 1.50 m/s

Variables	1.17 m/s	1.33 m/s	1.50 m/s	Prob.
Stride time (sec)	1.28 \pm 0.06	1.19 \pm 0.06 [†]	1.12 \pm 0.05 ^{††}	0.000 [*]
Stride length (m)	1.50 \pm 0.08	1.59 \pm 0.08 [†]	1.69 \pm 0.09 ^{††}	0.000 [*]
Stride frequency (strides/min)	47.0 \pm 2.39	50.5 \pm 2.53 [†]	53.5 \pm 2.50 ^{††}	0.000 [*]
Percentage of stride at toe-off (%)	65.2 \pm 1.73	64.5 \pm 1.78 [†]	63.6 \pm 1.64 ^{††}	0.000 [*]
Percentage of stride under double-support (%)	30.4 \pm 3.46	29.1 \pm 3.56 [†]	27.2 \pm 3.28 ^{††}	0.000 [*]

* statistically significant (p<.05) speed effect - ANOVA

† significantly (p<0.05) different from 1.17m/s - Tukey post-hoc test

‡ significantly (p<0.05) different from 1.33m/s - Tukey post-hoc test

Lower Body Sagittal Plane Ranges of Motion

Figures 4, 5, and 6 respectively depict angular motion around the ankle, knee, and hip at the three load carriage speeds. There were significant speed effects for maximum ankle angle, ankle range of motion, minimum hip angle, maximum hip angle, and hip range of motion (Table 3). There were no significant speed effects on minimum ankle angle, minimum knee angle, maximum knee angle, and knee range of motion. Maximum ankle angle, ankle range of motion, and maximum hip angle were significantly smaller at 1.17 m/s than at 1.33 and 1.50 m/s. Minimum hip angle decreased significantly and hip range of motion increased significantly with each speed increment.

The ankle was in an approximately neutral position at heel-strike, regardless of walking speed. Figure 4 shows that after heel-strike, the foot hit the ground at about the same speed and plantarflexed to the same degree regardless of walking speed. But the ankle subsequently dorsiflexed at an earlier percentage of stride as speed increased. However, the ultimate degree of dorsiflexion was the same at all speeds. Minimum ankle angle (maximal dorsiflexion) occurred at around heel-strike of the opposing foot, just as push-off forces peaked, with no inter-speed difference. After heel-strike of the opposing foot, the ankle plantarflexed sooner and to a greater degree as speed increased. Maximum ankle angle (maximum plantarflexion) occurred just before toe-off.

At heel-strike, knee angle averaged about 172° . The knee then flexed about 15° - 19° before it began to extend again at about the time of toe-off of the opposite foot, and kept extending until close to 50% of stride, when it was straighter on average than at heel-strike. It then started to flex as the contralateral foot approached the ground, continuing to do so until after heel-strike of the opposite foot and toe-off of the ipsilateral foot, reaching its minimum angle of about 115° during the rearward swing of the lower leg. As the leg was drawn forward the knee again began to extend, and continued doing so until just before heel-strike, when it began to flex again.

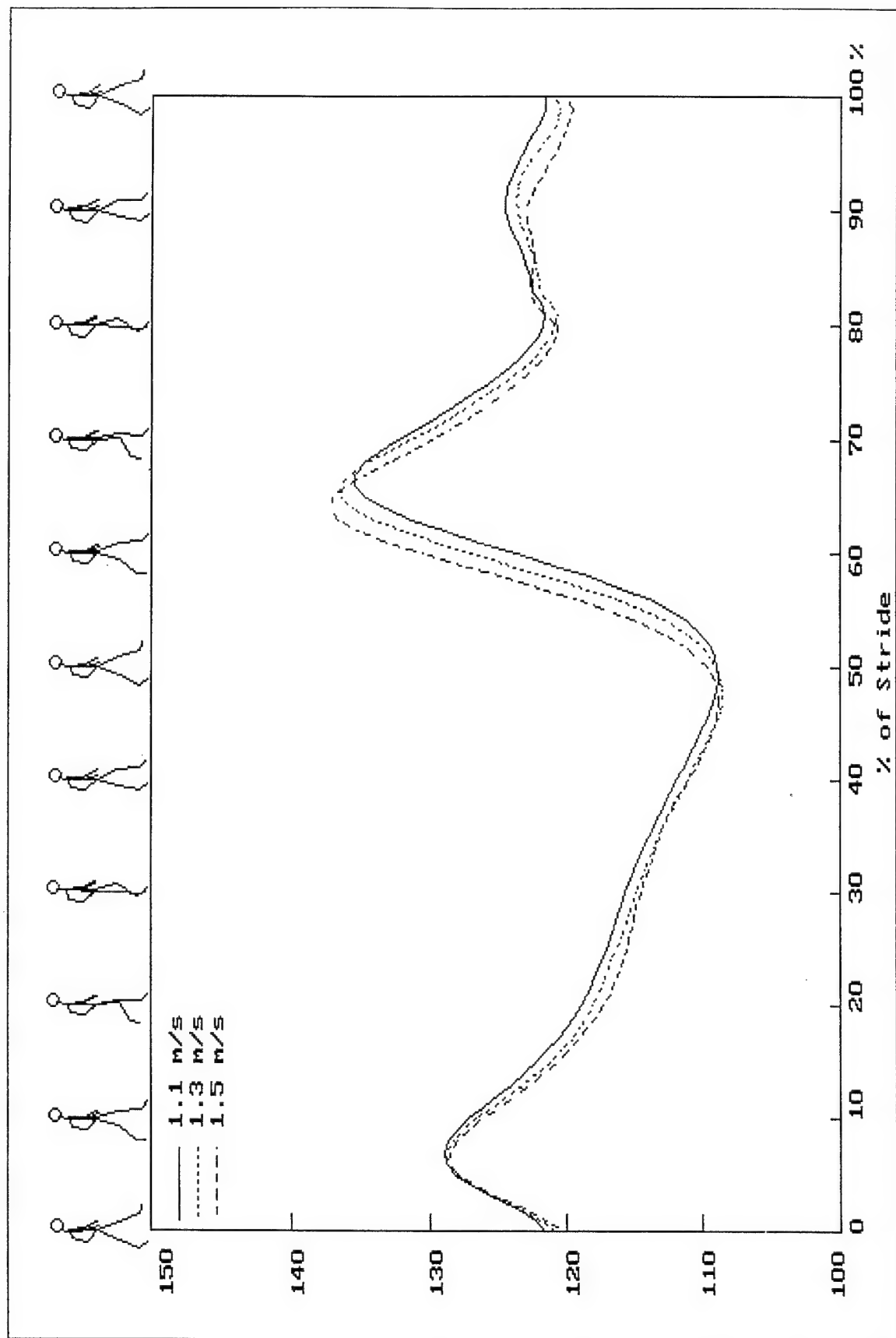


Figure 4. Speed effects for ankle angle.

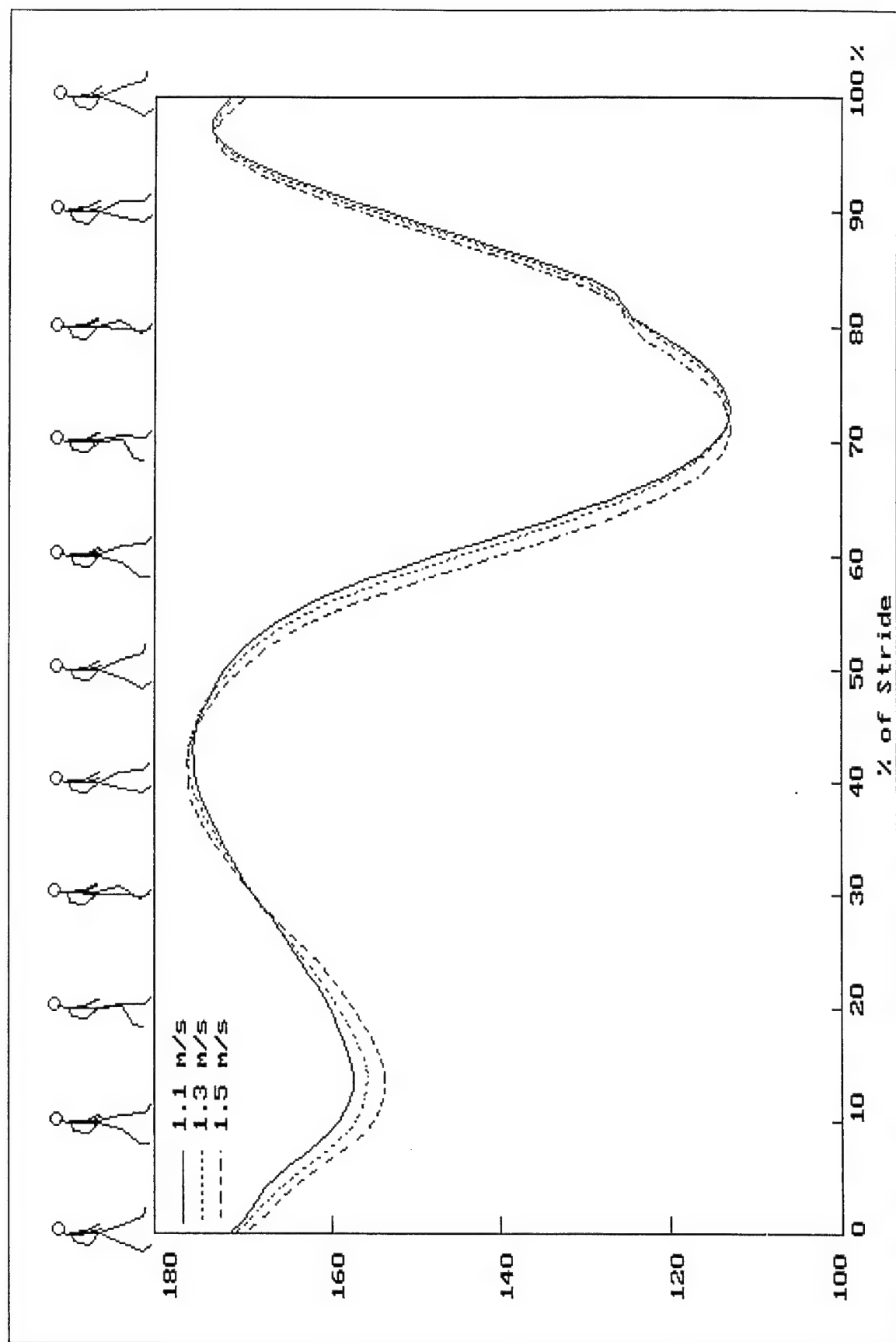


Figure 5. Speed effects for knee angle (degrees).

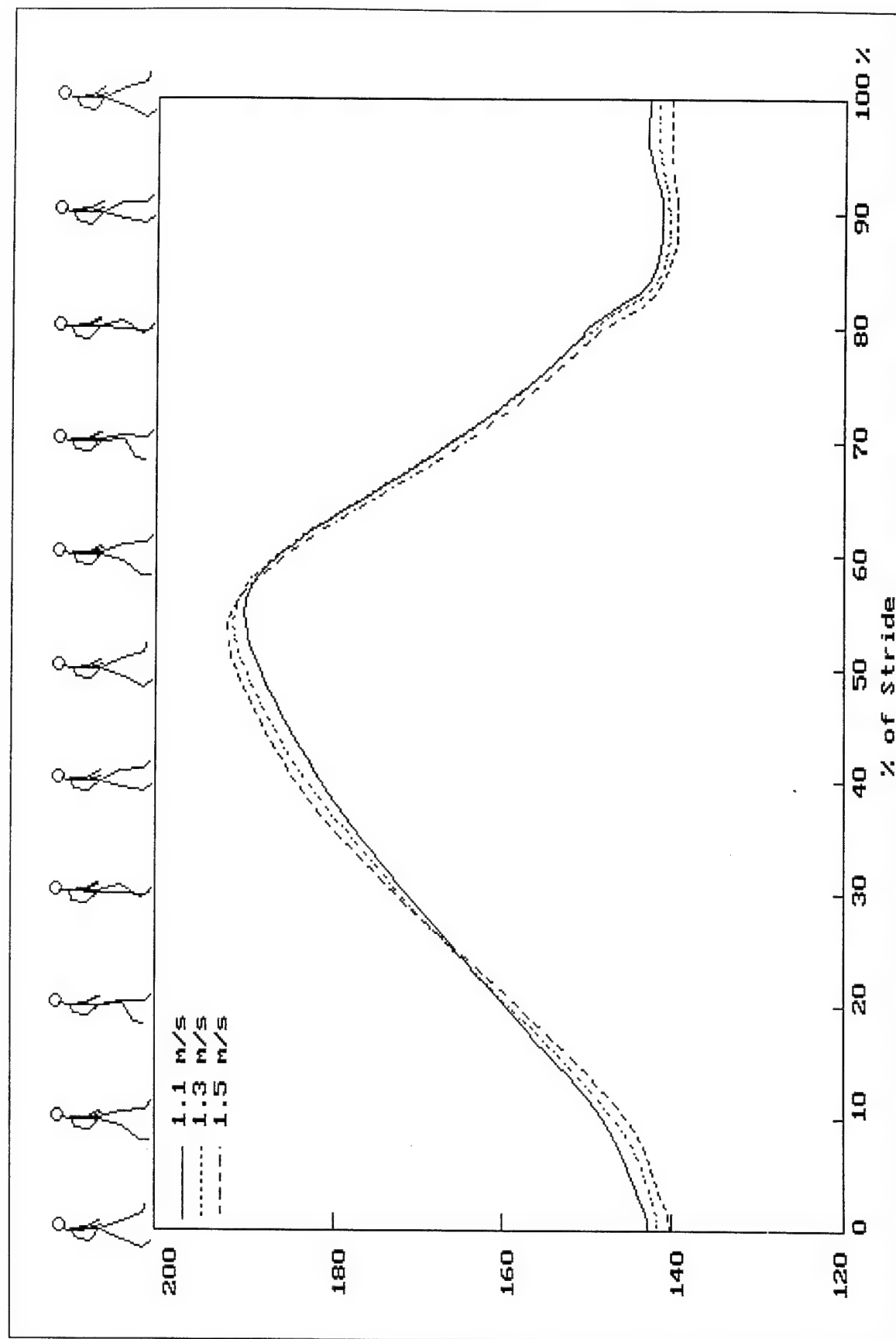


Figure 6. Speed effects for hip angle (degrees).

Table 3. Lower body sagittal plane ranges of motion (mean \pm SD) at load carriage speeds of 1.17, 1.33, and 1.50 m/s

Variables	1.17 m/s	1.33 m/s	1.50 m/s	Prob.
Minimum ankle angle (rad)	1.89 \pm 0.08	1.89 \pm 0.08	1.89 \pm 0.08	0.882
Maximum ankle angle (rad)	2.38 \pm 0.09	2.41 \pm 0.09 [†]	2.42 \pm 0.08 [†]	0.000 [*]
Ankle range of motion (rad)	0.49 \pm 0.05	0.52 \pm 0.05 [†]	0.54 \pm 0.07 [†]	0.000 [*]
Minimum knee angle (rad)	1.95 \pm 0.07	1.95 \pm 0.08	1.95 \pm 0.07	0.565
Maximum knee angle (rad)	3.10 \pm 0.09	3.10 \pm 0.09	3.10 \pm 0.09	0.647
Knee range of motion (rad)	1.14 \pm 0.11	1.15 \pm 0.10	1.15 \pm 0.09	0.502
Minimum hip angle (rad)	2.44 \pm 0.14	2.42 \pm 0.13 [†]	2.40 \pm 0.13 ^{†‡}	0.000 [*]
Maximum hip angle (rad)	3.32 \pm 0.12	3.34 \pm 0.12 [†]	3.36 \pm 0.11 [†]	0.000 [*]
Hip range of motion (rad)	0.88 \pm 0.09	0.92 \pm 0.09 [†]	0.95 \pm 0.08 ^{†‡}	0.000 [*]

Note - The joint angles are defined in Diagram.

* statistically significant ($p < 0.05$) speed effect - ANOVA

† significantly ($p < 0.05$) different from 1.17 m/s - Tukey post-hoc test

‡ significantly ($p < 0.05$) different from 1.33 m/s - Tukey post-hoc test

Upper Body Sagittal Plane Ranges of Motion

Minimum elbow angle, minimum shoulder angle, and shoulder range of motion were significantly affected by walking speed (Table 4). There were no significant speed effects on maximum elbow angle, elbow range of motion, maximum shoulder angle, minimum trunk angle, maximum trunk angle, and trunk range of motion. Minimum elbow angle at 1.17 m/s was significantly greater than at 1.33 and 1.50 m/s with no significant difference between 1.33 and 1.50 m/s. At 1.50 m/s, minimum shoulder angle was significantly smaller (indicating greater rearward arm swing), and shoulder range of motion was significantly greater than at 1.17 m/s with no other significant inter-speed differences.

Table 4. Upper body sagittal plane ranges of motion (mean \pm SD) at load carriage speeds of 1.17, 1.33, and 1.50 m/s

Variables	1.17 m/s	1.33 m/s	1.50 m/s	Prob.
Minimum elbow angle (rad)	2.54 \pm 0.11	2.50 \pm 0.11 [†]	2.48 \pm 0.10 [†]	0.002*
Maximum elbow angle (rad)	2.81 \pm 0.14	2.82 \pm 0.11	2.83 \pm 0.12	0.323
Elbow range of motion (rad)	0.29 \pm 0.25	0.30 \pm 0.13	0.35 \pm 0.14	0.160
Minimum shoulder angle (rad)	-0.11 \pm 0.11	-0.12 \pm 0.12	-0.14 \pm 0.12 [†]	0.018*
Maximum shoulder angle (rad)	0.22 \pm 0.19	0.25 \pm 0.16	0.27 \pm 0.17	0.072
Shoulder range of motion (rad)	0.34 \pm 0.24	0.37 \pm 0.20	0.42 \pm 0.24 [†]	0.014*
Minimum trunk angle (rad)	1.38 \pm 0.10	1.38 \pm 0.10	1.38 \pm 0.10	0.193
Maximum trunk angle (rad)	1.46 \pm 0.10	1.45 \pm 0.09	1.45 \pm 0.09	0.241
Trunk range of motion (rad)	0.07 \pm 0.02	0.07 \pm 0.02	0.07 \pm 0.02	0.921

* statistically significant ($p < 0.05$) speed effect - ANOVA

[†] significantly ($p < 0.05$) different from 1.17 m/s - Tukey post-hoc test

[‡] significantly ($p < 0.05$) different from 1.33 m/s - Tukey post-hoc test

Center of Mass Parameters

Both the minimum and maximum horizontal center of mass velocities, as well as the absolute difference between them increased significantly with load carriage speed (Table 5). Yet the ratio between the minimum and maximum horizontal center of mass velocities stayed constant at 0.54.

Upward and downward maximum center of mass velocities, minimum vertical position, and vertical range of motion were significantly affected by speed, while maximal vertical position was not affected. Maximum upward velocity was significantly greater at 1.50 m/s than at 1.17 m/s with no other significant inter-speed differences. Maximum downward velocity increased significantly with each increment in walking speed. At the 1.17 m/s speed, maximum upward velocity was higher than maximum downward velocity, but at the highest speed, maximum downward velocity was greater. Minimum vertical position at the 1.50 m/s speed was significantly lower than at 1.17 and 1.33 m/s with no significant difference between 1.17 and 1.33 m/s. Vertical range of motion increased significantly with each speed increment. Figure 7 shows how the increase in vertical range of motion was due completely to lower minimum vertical position with increasing speed; maximum vertical position did not change with speed.

Table 5. Center of mass parameters (mean \pm SD) at load carriage speeds of 1.17, 1.33, and 1.50 m/s

Variables	1.17 m/s	1.33 m/s	1.50 m/s	Prob.
Maximum horizontal velocity (m/sec)	1.49 \pm 0.13	1.65 \pm 0.10 [†]	1.85 \pm 0.13 ^{††}	0.000*
Minimum horizontal velocity (m/sec)	0.80 \pm 0.10	0.90 \pm 0.13 [†]	1.00 \pm 0.16 ^{††}	0.000*
Maximum upward velocity (m/sec)	0.32 \pm 0.10	0.34 \pm 0.14	0.36 \pm 0.07 [†]	0.022*
Maximum downward velocity (m/sec)	0.27 \pm 0.06	0.33 \pm 0.09 [†]	0.38 \pm 0.08 ^{††}	0.000*
Maximum vertical position (m)	0.99 \pm 0.06	0.99 \pm 0.07	0.99 \pm 0.07	0.570
Minimum vertical position (m)	0.95 \pm 0.06	0.94 \pm 0.06	0.93 \pm 0.06 ^{††}	0.000*
Vertical range of position (m)	0.04 \pm 0.01	0.05 \pm 0.01 [†]	0.06 \pm 0.01 ^{††}	0.000*

* statistically significant ($p < .05$) speed effect - ANOVA

† significantly ($p < 0.05$) different from 1.17 m/s - Tukey post-hoc test

‡ significantly ($p < 0.05$) different from 1.33 m/s - Tukey post-hoc test

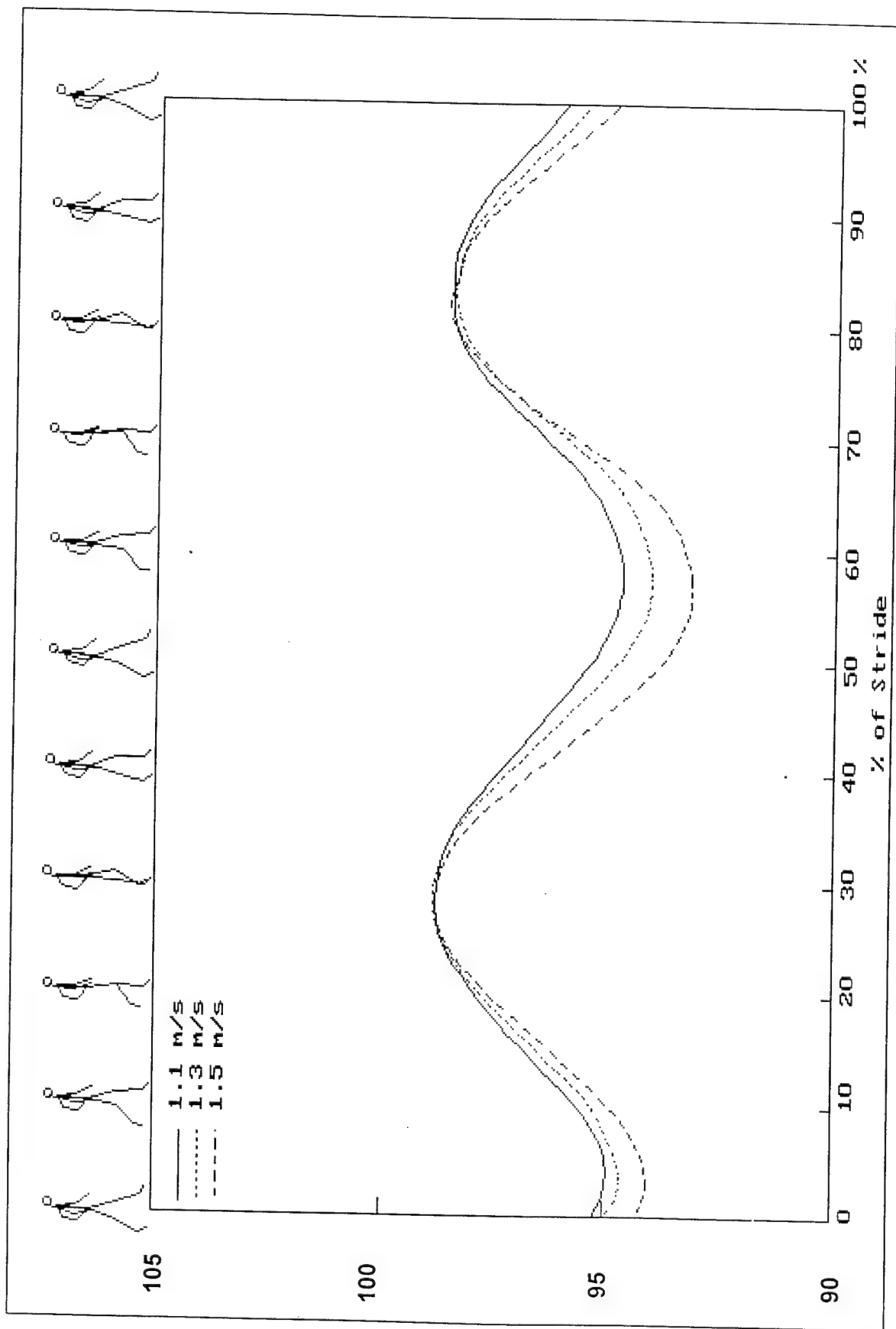


Figure 7. Speed effects for center of mass vertical position (cm).

Ground Reaction Forces and Impulses

Vertical, front-back, and medio-lateral ground reaction forces are respectively depicted in Figures 8, 9, and 10, all of which show generally greater forces with increasing speed. There were significant speed effects for braking impulse, medial impulse, vertical impulse, average propulsive force, and average braking force (Table 6). Not significantly affected by speed were propulsive impulse, lateral impulse, average lateral force, average medial force, and average vertical force. Braking impulse was significantly greater and medial impulse decreased significantly as speed increased, with significant no other significant inter-speed differences. Vertical impulse decreased significantly as speed increased, with significant differences among all speeds. Average propulsive force and average braking force both increased significantly with each increment in speed. When examining Figures 8-10, it is important to remember that the x-axis is percentage of stride rather than time. Thus, while the patterns align well in terms of percentage of stride, the stride pattern at the higher walking speed actually occurs over a shorter period of time. That explains why, even though the force magnitudes were generally higher at the faster speeds than at the slower speeds, the impulses (force•time) were generally lower.

Table 6. Ground reaction impulses and forces (mean \pm SD) at load carriage speeds of 1.17, 1.33, and 1.50 m/s

Variables	1.17 m/s	1.33 m/s	1.50 m/s	Prob.
Propulsive impulse (N•s)	34.7 \pm 7.33	35.7 \pm 7.15	36.1 \pm 7.99	0.139
Braking impulse (N•s)	34.1 \pm 7.87	35.2 \pm 7.42	36.7 \pm 7.79 [†]	0.002*
Lateral impulse (N•s)	1.60 \pm 1.02	1.70 \pm 1.09	1.86 \pm 1.20	0.063
Medial impulse (N•s)	21.6 \pm 7.08	20.2 \pm 8.16	19.1 \pm 7.80 [†]	0.013*
Vertical impulse (N•s)	626 \pm 107	582 \pm 99 [†]	548 \pm 97 ^{†‡}	0.000*
Average propulsive force (N)	78.7 \pm 18.3	87.7 \pm 17.6 [†]	95.3 \pm 20.1 ^{†‡}	0.000*
Average braking force (N)	82.0 \pm 17.9	90.4 \pm 18.1 [†]	99.0 \pm 18.6 ^{†‡}	0.000*
Average lateral force (N)	11.9 \pm 5.60	12.4 \pm 5.08	13.7 \pm 5.30	0.069
Average medial force (N)	29.3 \pm 8.4	30.0 \pm 10.5	30.5 \pm 10.4	0.508
Average vertical force (N)	724 \pm 118	726 \pm 113	729 \pm 113	0.059

* statistically significant ($p < .05$) speed effect - ANOVA

† significantly ($p < .05$) different from 1.17 m/s - Tukey post-hoc test

‡ significantly ($p < .05$) different from 1.33 m/s - Tukey post-hoc test

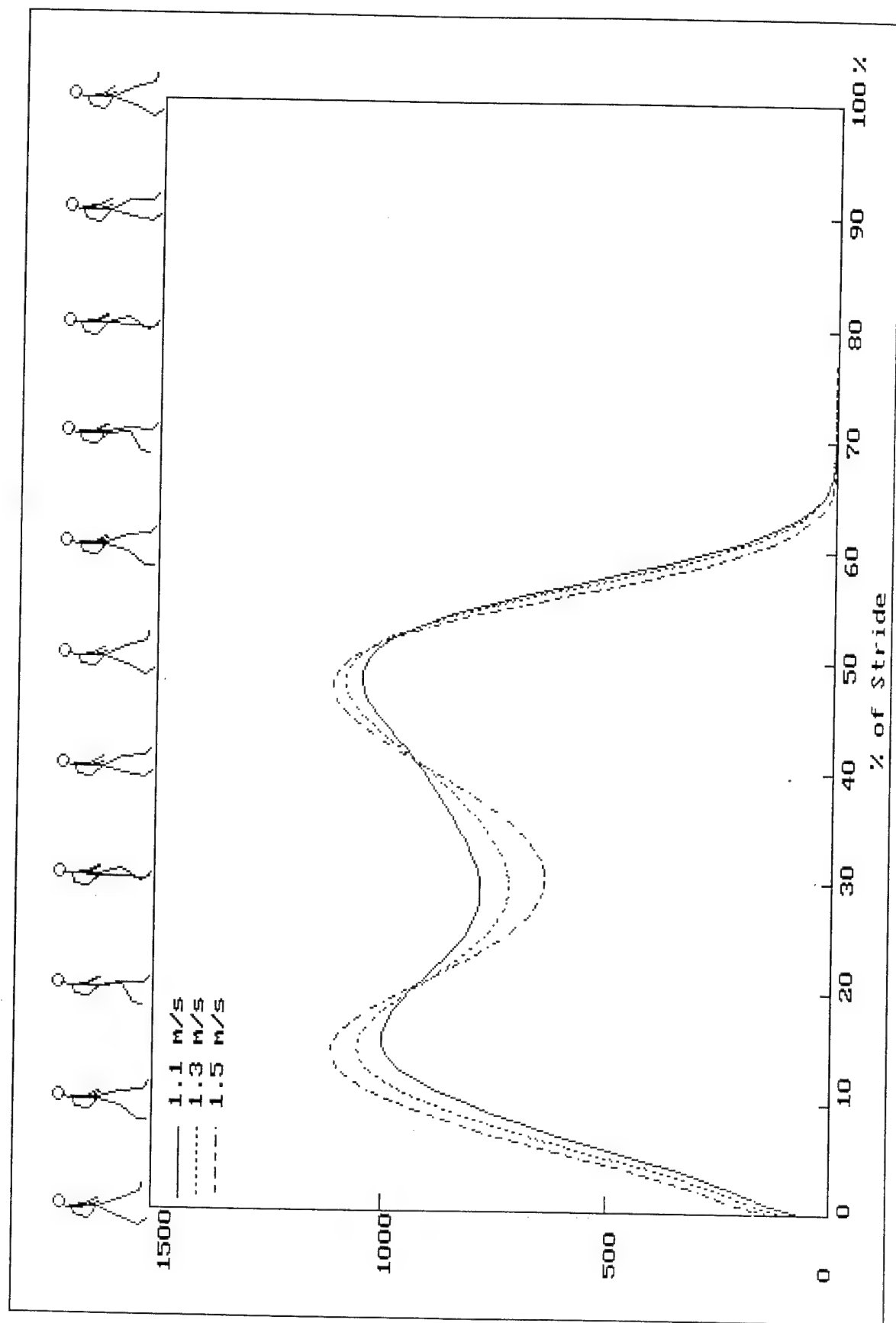


Figure 8. Speed effects for vertical ground reaction force (N).

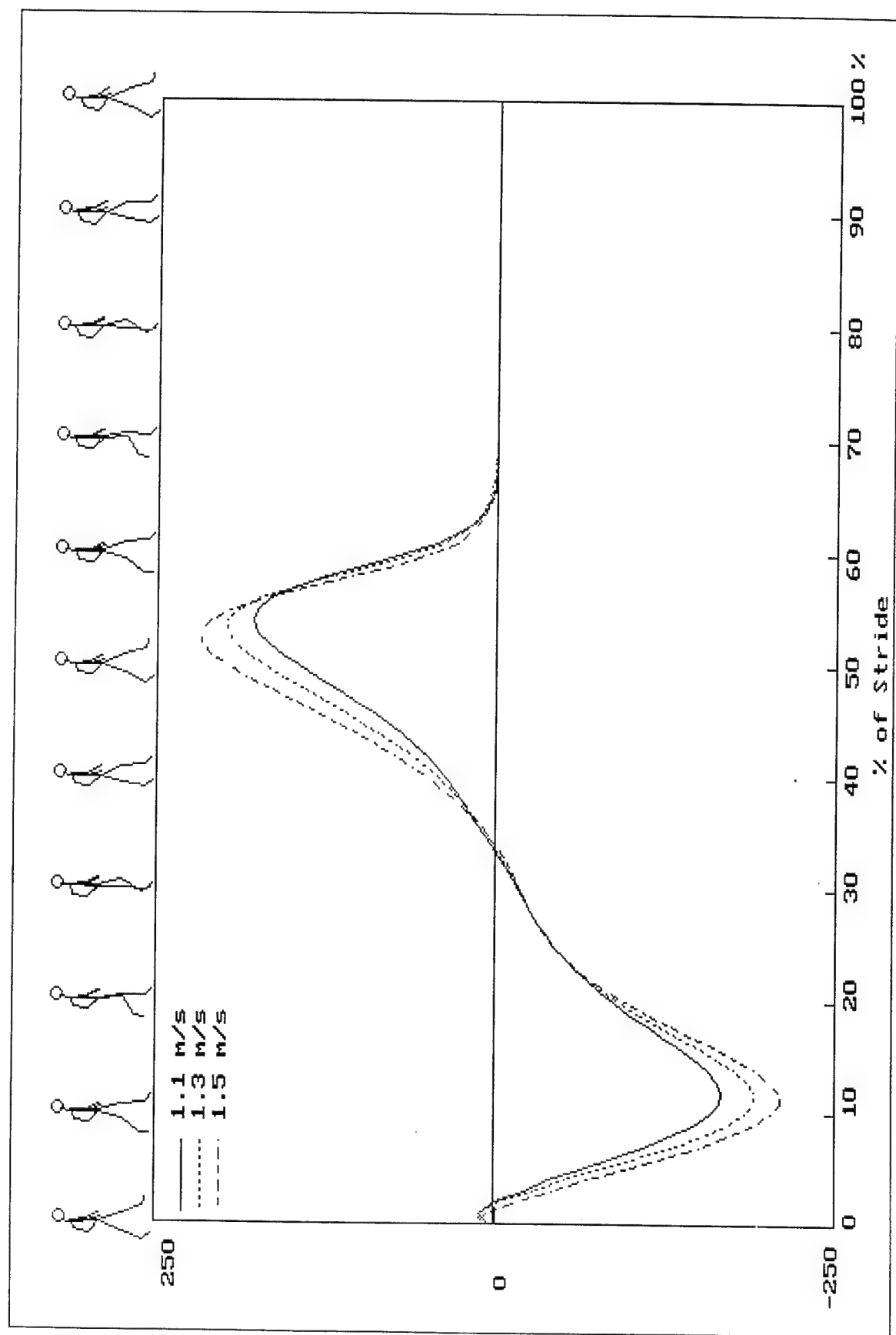


Figure 9. Speed effects for front-back ground reaction force (N).

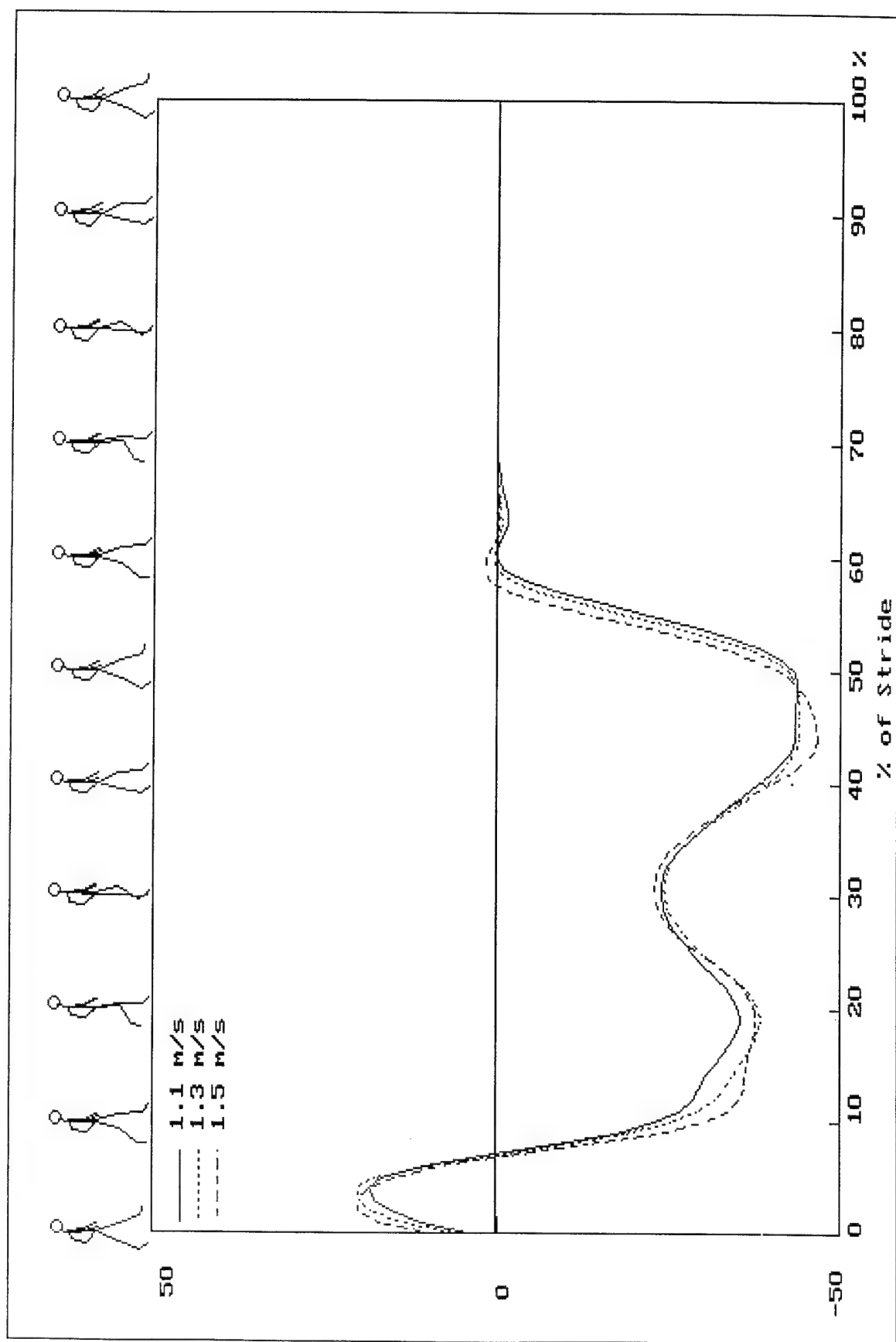


Figure 10. Speed effects for medio-lateral ground reaction force (N).

Peak propulsive force, peak braking force, peak lateral force, first peak vertical force, and second peak vertical force were significantly affected by walking speed (Table 7). Peak propulsive and braking forces as well as the first and second peak vertical forces significantly increased with each increment in speed. Peak lateral force at the 1.50 m/s speed was significantly greater than at the 1.17 m/s speed, with no other significant inter-speed differences.

As to percentage of stride at which peak forces occurred, there were no significant speed effects for peak braking, lateral, medial, and second peak vertical forces. However, peak propulsive force occurred at a significantly lesser percentage of stride at 1.50 m/s than 1.17 and 1.33 m/s, with no other significant inter-speed differences. The first peak occurred at a significantly lesser percentage of stride at 1.50 m/s than at 1.17 m/s, with no other significant inter-speed differences. There was the same degree of increase in peak braking force when walking speed increased from 1.17 to 1.33 m/s as when walking speed increased from 1.33 to 1.50. The same pattern of increase occurred for peak propulsive force as well as for average braking and propulsive force.

Table 7. Peak ground reaction forces and timing (mean \pm SD) at load carriage speeds of 1.17, 1.33, and 1.50 m/s

Variables	1.17 m/s	1.33 m/s	1.50 m/s	Prob.
Peak propulsive force (PPF) (N)	186 \pm 37.0	208 \pm 37.8 [†]	228 \pm 43.7 ^{††}	0.000*
% of stride at PPF	55.1 \pm 1.72	54.7 \pm 1.51	54.0 \pm 1.59 ^{††}	0.000*
Peak braking force (PBF) (N)	175 \pm 40.7	197 \pm 42.4 [†]	220 \pm 44.8 ^{††}	0.000*
% of stride at PBF	12.3 \pm 1.47	12.4 \pm 1.11	12.3 \pm 1.08	0.786
Peak lateral force (PLF) (N)	24.0 \pm 12.5	26.2 \pm 11.6	29.1 \pm 13.3 [†]	0.001*
% of stride at PLF	4.40 \pm 1.44	4.25 \pm 1.64	4.03 \pm 1.75	0.288
Peak medial force (PMF) (N)	50.3 \pm 12.8	52.1 \pm 17.1	54.2 \pm 17.7	0.187
% of stride at PMF	38.3 \pm 14.2	37.5 \pm 14.1	36.8 \pm 13.9	0.698
1st Peak vertical force (PVF1) (N)	1,015 \pm 176	1,064 \pm 176 [†]	1,128 \pm 188 ^{††}	0.000*
% of stride at PVF1	16.3 \pm 2.19	15.6 \pm 1.34	15.2 \pm 1.11 [†]	0.003*
2nd Peak vertical force (PVF2) (N)	1,061 \pm 175	1,097 \pm 184 [†]	1,125 \pm 188 ^{††}	0.000*
% of stride at PVF2	48.9 \pm 1.77	48.9 \pm 1.49	48.6 \pm 1.33	0.238

* statistically significant ($p < .05$) speed effect - ANOVA

† significantly ($p < .05$) different from 1.17 m/s - Tukey post-hoc test

‡ significantly ($p < .05$) different from 1.33 m/s - Tukey post-hoc test

Joint Forces and Torques

Figures 11, 12, and 13 respectively depict torque about the ankle, knee, and hip during a full stride. There were significant speed effects on peaks of the following variables: forward, backward, upward, and downward shank-on-foot and thigh-on-shank forces; forward, backward, and downward trunk-on-thigh forces; knee and hip flexion and extension torques, and ankle dorsiflexion torque (Table 8). There were no significant inter-speed differences for upward trunk-on-hip force, ankle plantarflexion torque, and knee flexion torque. For all but three of the variables showing significant speed effects, means increased significantly with each increment in speed. The exceptions included peak upward shank-on-foot force and peak backward trunk-on-hip force which both showed significant differences only between 1.17 m/s and the two faster speeds, as well as peak ankle dorsiflexion torque, which showed significant differences only between 1.50 m/s and the two slower speeds.

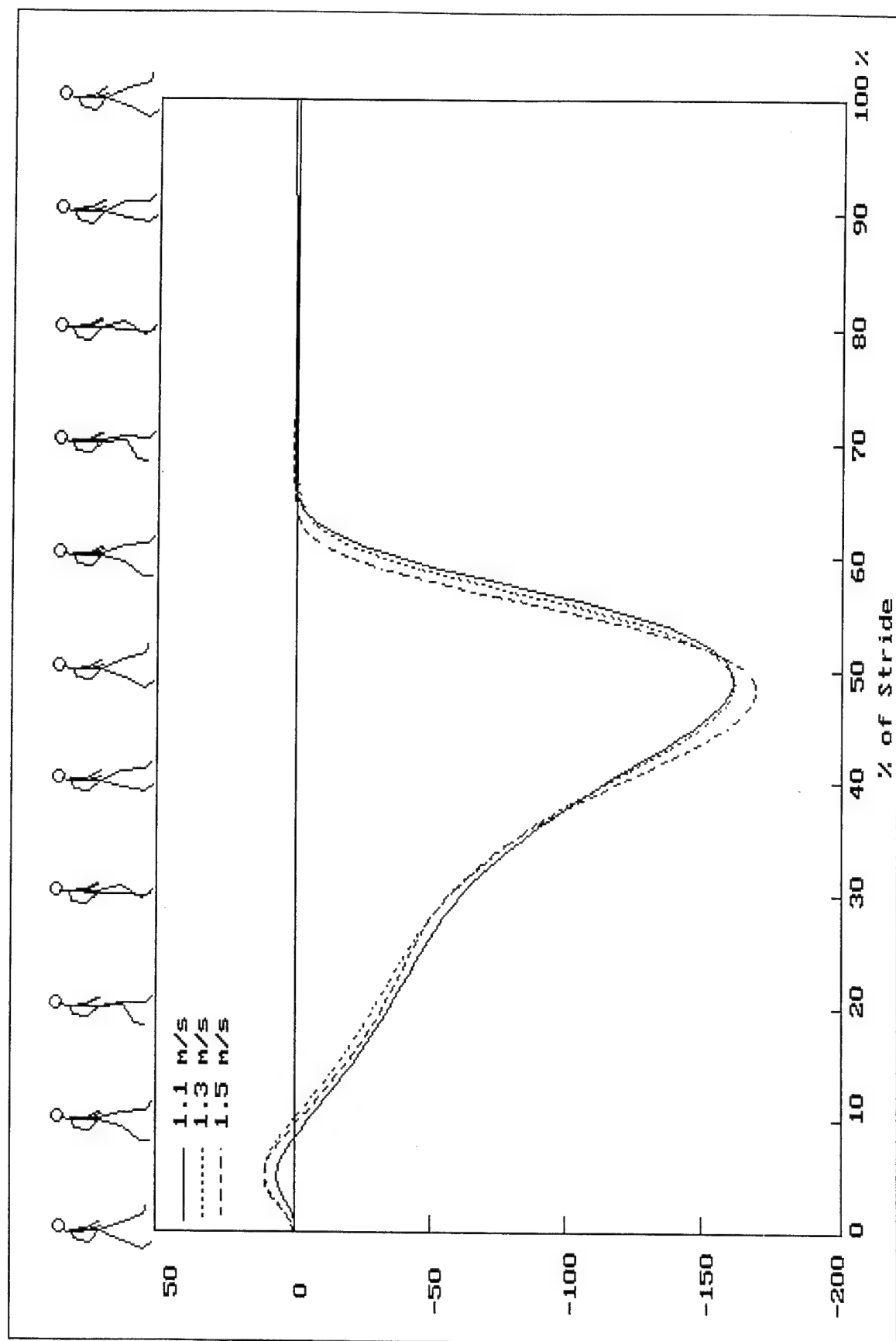


Figure 11. Speed effects for ankle torque (N.m).

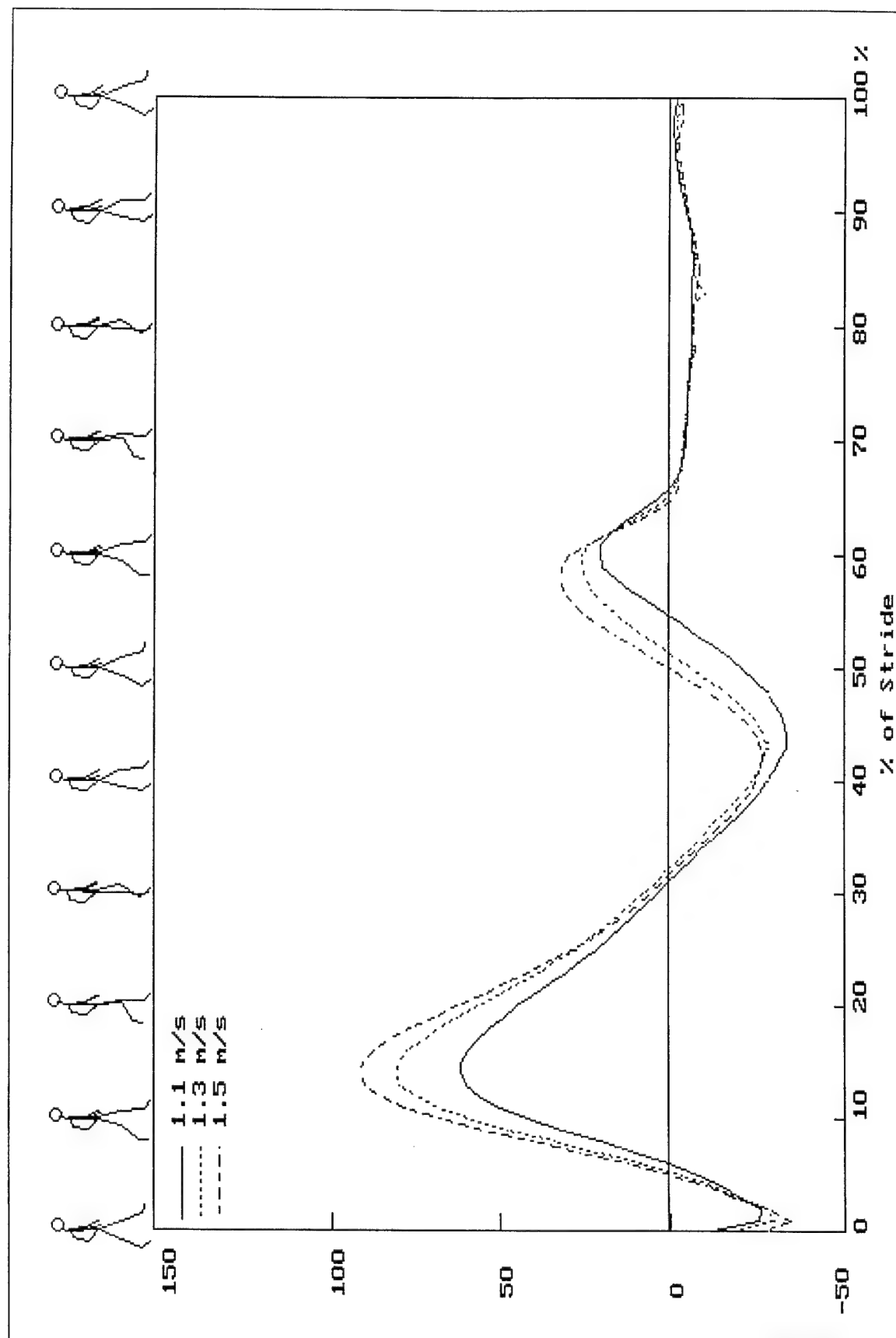


Figure 12. Speed effects for knee torque (N.m).

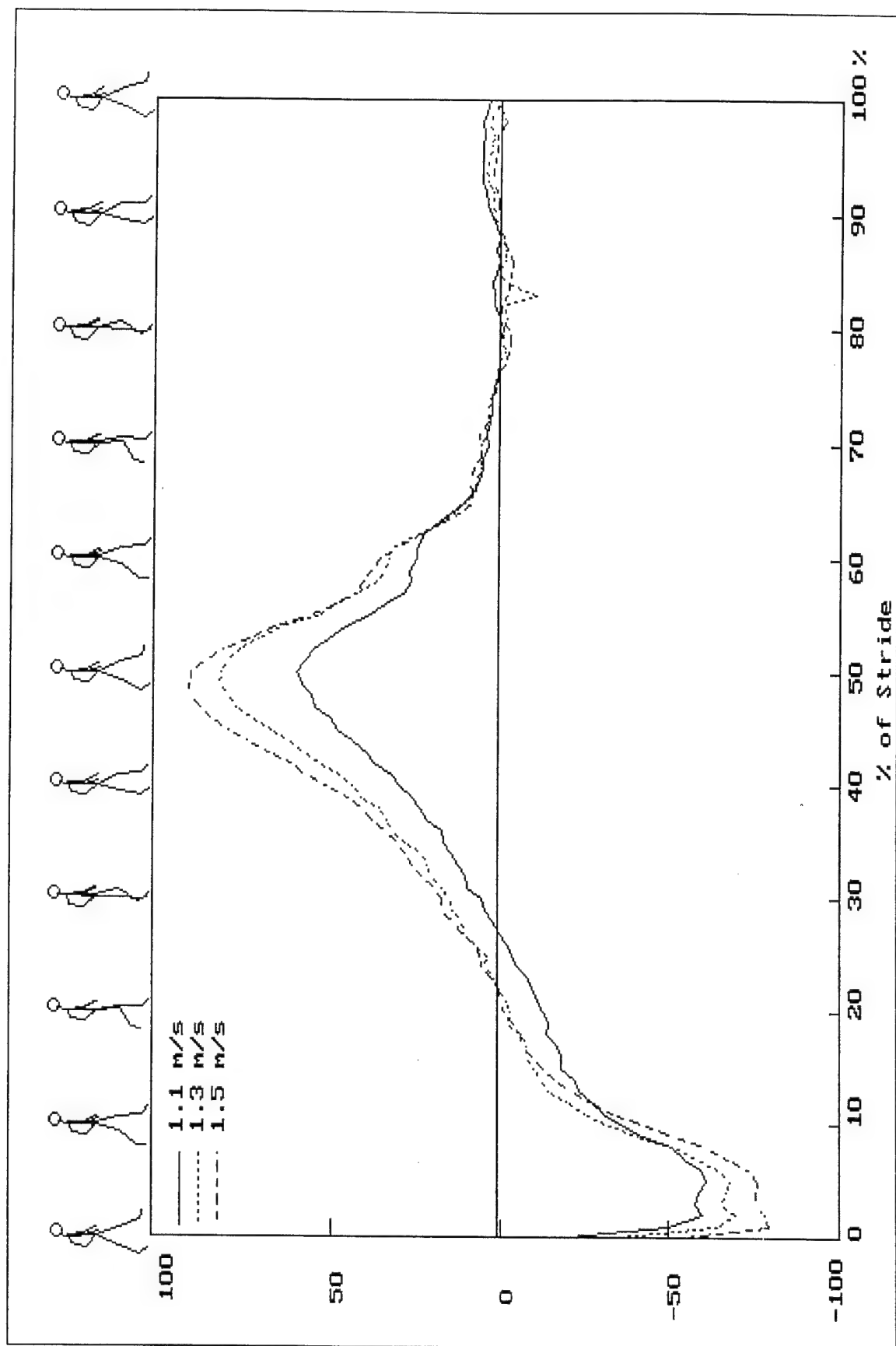


Figure 13. Speed effects for hip torque (N.m).

Table 8. Joint forces and torques (mean \pm SD) at load carriage speeds of 1.17, 1.33, and 1.50 m/s

Variables	1.17 m/s	1.33 m/s	1.50 m/s	Prob.
peak forward shank-on-foot force (N)	172 \pm 40.1	193 \pm 42.2 [†]	213 \pm 44.2 ^{††}	0.000*
peak backward shank-on-foot force (N)	179 \pm 36.4	199 \pm 36.9 [†]	217 \pm 42.6 ^{††}	0.000*
peak upward shank-on-foot force (N)	14.1 \pm 1.82	14.6 \pm 1.91 [†]	14.7 \pm 1.90 [†]	0.000*
peak downward shank-on-foot force (N)	1,064 \pm 181	1,110 \pm 189 [†]	1,159 \pm 198 ^{††}	0.000*
peak ankle dorsiflexion torque (N•m)	8.13 \pm 5.43	8.11 \pm 5.35	10.88 \pm 6.54 ^{††}	0.005*
peak ankle plantarflexion torque(N•m)	164 \pm 33.4	170 \pm 41.0	173 \pm 33.7	0.099
peak forward thigh-on-shank force (N)	162 \pm 39.0	180 \pm 40.7 [†]	199 \pm 43.6 ^{††}	0.000*
peak backward thigh-on-shank force (N)	160 \pm 34.7	177 \pm 35.6 [†]	191 \pm 40.8 ^{††}	0.000*
peak upward thigh-on-shank force (N)	55.2 \pm 6.95	56.6 \pm 6.99	55.9 \pm 7.77	0.037*
peak downward thigh-on-shank force (N)	1,029 \pm 180	1,073 \pm 196 [†]	1,122 \pm 196 ^{††}	0.000*
peak knee extension torque (N•m)	62.9 \pm 23.2	76.9 \pm 30.7 [†]	92.7 \pm 38.9 ^{††}	0.000*
peak knee flexion torque (N•m)	44.9 \pm 18.4	45.4 \pm 19.6	46.0 \pm 15.8	0.993
peak forward trunk-on-hip force (N)	155 \pm 39.3	177 \pm 45.7 [†]	192 \pm 48.3 ^{††}	0.000*
peak backward trunk-on-hip force (N)	151 \pm 34.3	167 \pm 36.2 [†]	175 \pm 40.1 [†]	0.000*
peak upward trunk-on-hip force (N)	147 \pm 27.8	177 \pm 45.7	192 \pm 48.3	0.743
peak downward trunk-on-hip force (N)	999 \pm 184	1,042 \pm 187 [†]	1,090 \pm 198 ^{††}	0.000*
peak hip extension torque (N•m)	64.8 \pm 26.5	82.5 \pm 41.6 [†]	98.4 \pm 43.5 ^{††}	0.000*
peak hip flexion torque (N•m)	72.9 \pm 19.9	88.6 \pm 27.9 [†]	100.3 \pm 26.6 ^{††}	0.000*

* statistically significant (p<.05) speed effect - ANOVA

† significantly (p<0.05) different from 1.17 m/s - Tukey post-hoc test

‡ significantly (p<0.05) different from 1.33 m/s - Tukey post-hoc test

Muscle Electrical Activity

There were significant speed effects on average muscle electrical activity of all muscles monitored, including the trapezius, spinal erector, quadriceps, hamstrings, tibialis anterior, and gastrocnemius (Table 9). While means for muscle activities increased with walking speed, the differences between 1.17 and 1.33 m/s were not significant. For the trapezius, quadriceps, and tibialis anterior, electrical activity at 1.50 m/s was only significantly different from that at 1.17 m/s. For the spinal erectors, hamstrings, and gastrocnemius, electrical activity at 1.50 m/s was significantly greater than at both 1.17 and 1.33 m/s.

Table 9. Muscle electrical activity (mean \pm SD) at load carriage speeds of 1.17, 1.33, and 1.50 m/s

Variables	1.17 m/s	1.33 m/s	1.50 m/s	Prob.
Average trapezius EMG (μ V)	42.8 \pm 28.8	46.6 \pm 29.5	50.2 \pm 29.3 [†]	0.029*
Average spinal erector EMG (μ V)	75.2 \pm 51.7	76.6 \pm 39.1	92.5 \pm 53.0 ^{†‡}	0.009*
Average quadriceps EMG (μ V)	77.5 \pm 56.5	96.8 \pm 70.0	108.5 \pm 75.3 [†]	0.002*
Average hamstrings EMG (μ V)	77.2 \pm 35.5	92.7 \pm 43.2	141.4 \pm 129.0 ^{†‡}	0.000*
Average tibialis anterior EMG (μ V)	242 \pm 109	274 \pm 121	315 \pm 140 [†]	0.003*
Average gastrocnemius EMG (μ V)	189 \pm 103	206 \pm 104	242 \pm 123 ^{†‡}	0.000*

Note - Table values are EMG amplitudes averaged over one full stride.

* statistically significant ($p < .05$) speed effect - ANOVA

† significantly ($p < 0.05$) different from 1.17 m/s - Tukey post-hoc test

‡ significantly ($p < 0.05$) different from 1.33 m/s - Tukey post-hoc test

Backpack Acceleration

Peak upward, downward, and backward backpack accelerations increased significantly with each increment in walking speed, while peak forward acceleration showed no significant effect (Table 10).

Table 10. Backpack accelerations (mean \pm SD) at load carriage speeds of 1.17, 1.33, and 1.50 m/s

Variables	1.17 m/s	1.33 m/s	1.50 m/s	Prob.
Peak upward acceleration (g)	0.36 \pm 0.18	0.42 \pm 0.08 [†]	0.48 \pm 0.10 ^{††}	0.000 [*]
Peak downward acceleration (g)	0.27 \pm 0.06	0.34 \pm 0.07 [†]	0.44 \pm 0.09 ^{††}	0.000 [*]
Peak backward acceleration (g)	0.53 \pm 0.17	0.59 \pm 0.19 [†]	0.65 \pm 0.21 ^{††}	0.000 [*]
Peak forward acceleration (g)	0.06 \pm 0.09	0.08 \pm 0.11	0.06 \pm 0.09	0.058

* statistically significant ($p < 0.05$) speed effect - ANOVA

† significantly ($p < 0.05$) different from 1.17 m/s - Tukey post-hoc test

‡ significantly ($p < 0.05$) different from 1.33 m/s - Tukey post-hoc test

LOAD-SPEED INTERACTION EFFECTS

Significant load-speed interaction effects were found for stride time, stride frequency, vertical ground reaction impulse, downward shank-on-foot impulse, downward thigh-on-shank impulse, and downward trunk-on-thigh impulse (Figures 14-19). The interaction for stride time (Figure 14) was apparently due to the fact that stride time was fairly constant over the lowest three loads at all speeds, but dropped off for the highest load at the 1.17 and 1.33 m/s speeds but not at 1.50 m/s. Since stride frequency (Figure 15) was calculated as $60/(\text{stride time in seconds})$, the two variables showed similar interactions. Stride frequency was fairly constant over the lowest three loads but rose for the highest load at the 1.17 and 1.33 m/s speeds but not at 1.50 m/s.

The load-speed interaction for the vertical ground reaction impulse, downward shank-on-foot, thigh-on-shank, and trunk-on-thigh impulses are similar because forces at the foot are transmitted through the ankle, knee, and hip. It should be noted that, while the patterns in Figures 16-19 appear very similar, the magnitudes of the impulses get lower from the ground to the ankle (shank-on-foot) to the knee (thigh-on-shank) to the hip (trunk-on-thigh), as indicated by reduction in the y-axis scale. However, all these impulses increased consistently as the load increased for walking speeds of 1.17 and 1.50 m/s speeds. The interaction resulted from the fact that a somewhat different pattern emerged for the 1.33 m/s speed, at which impulse did not increase as much when going from the next-highest to the highest load as it did at the other walking speeds.

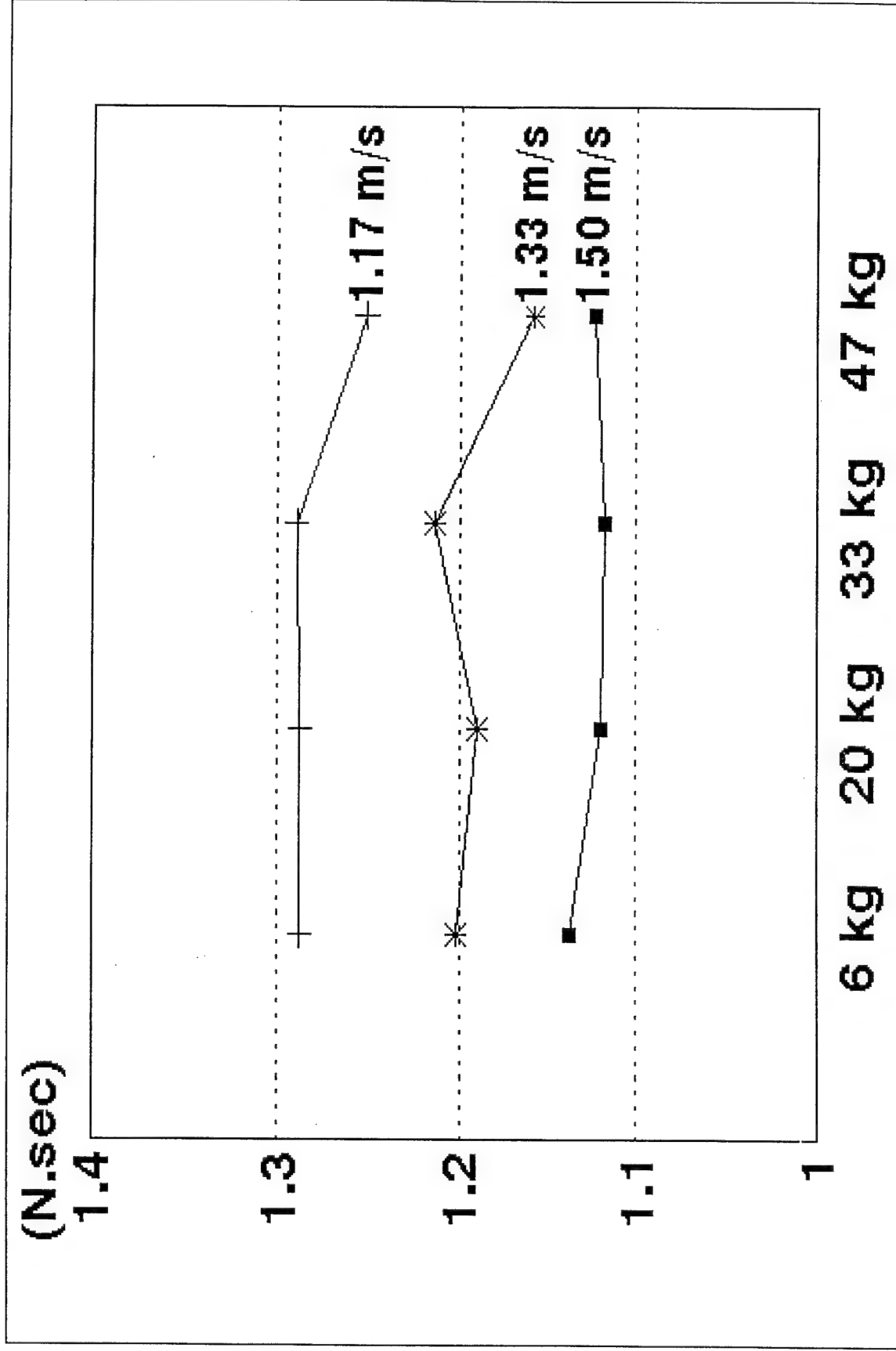


Figure 14. Interaction effects for stride time.

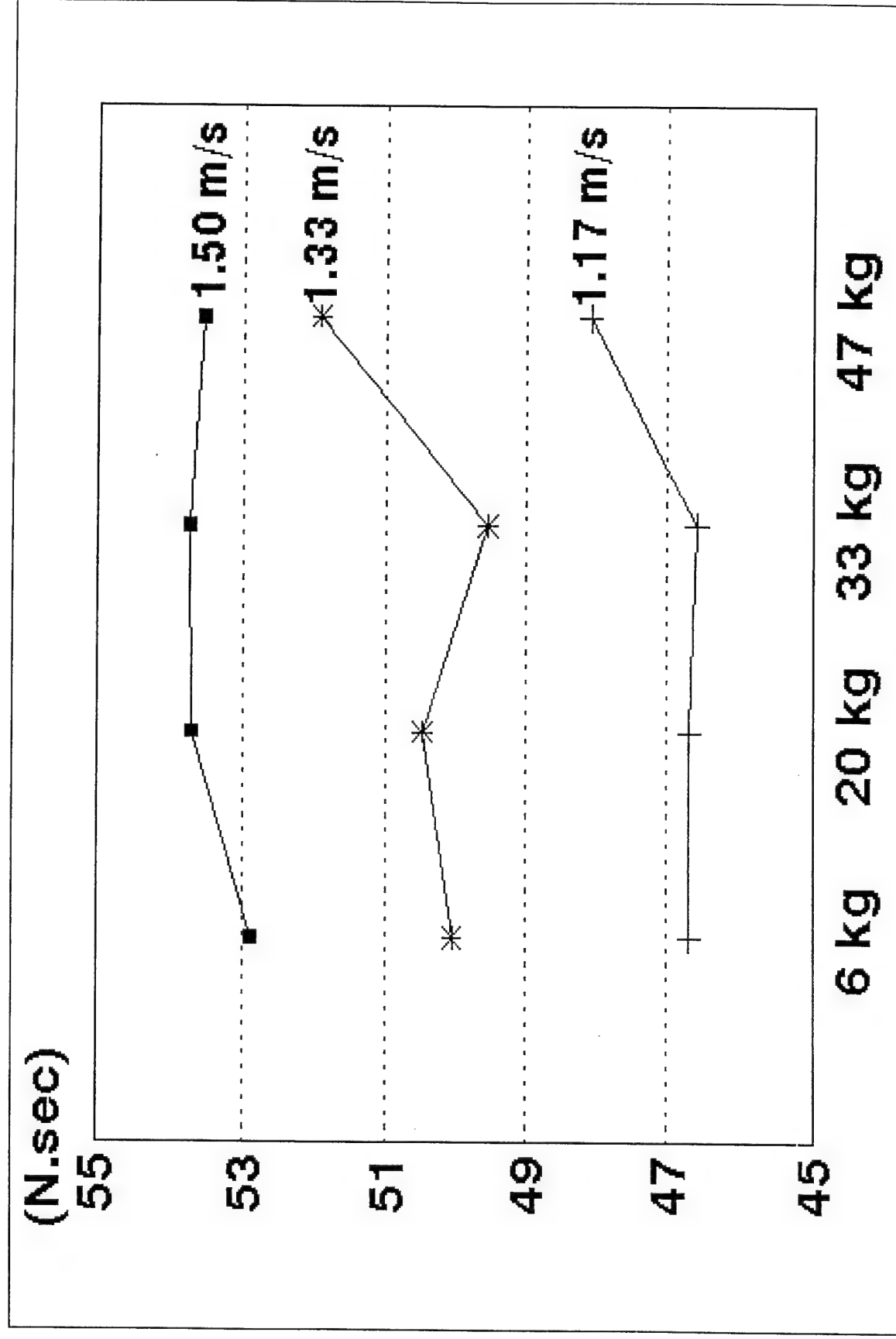


Figure 15. Interaction effects for stride frequency.

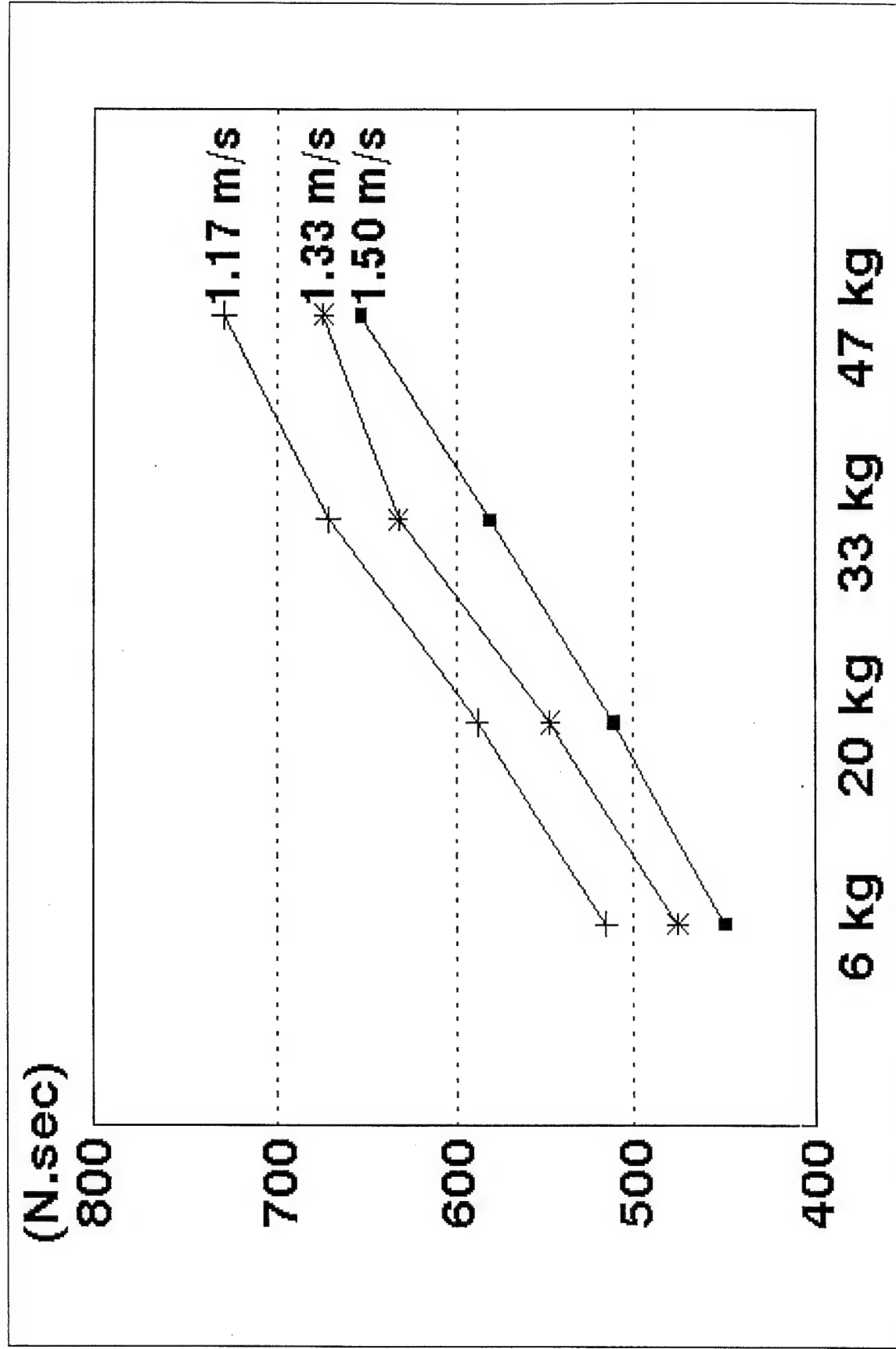


Figure 16. Interaction effects for vertical ground reaction impulse.

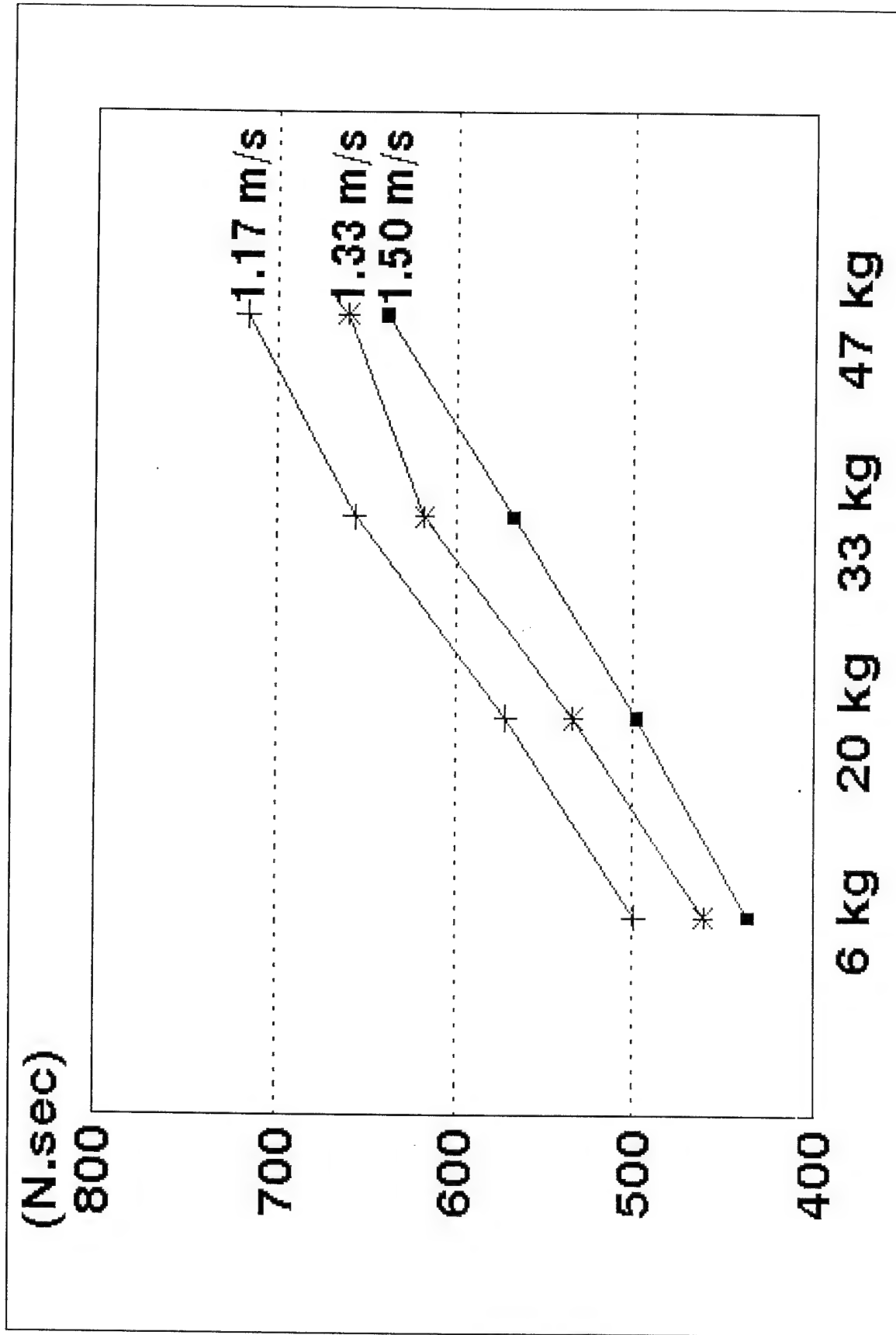


Figure 17. Interaction effects for downward shank-on-foot impulse.

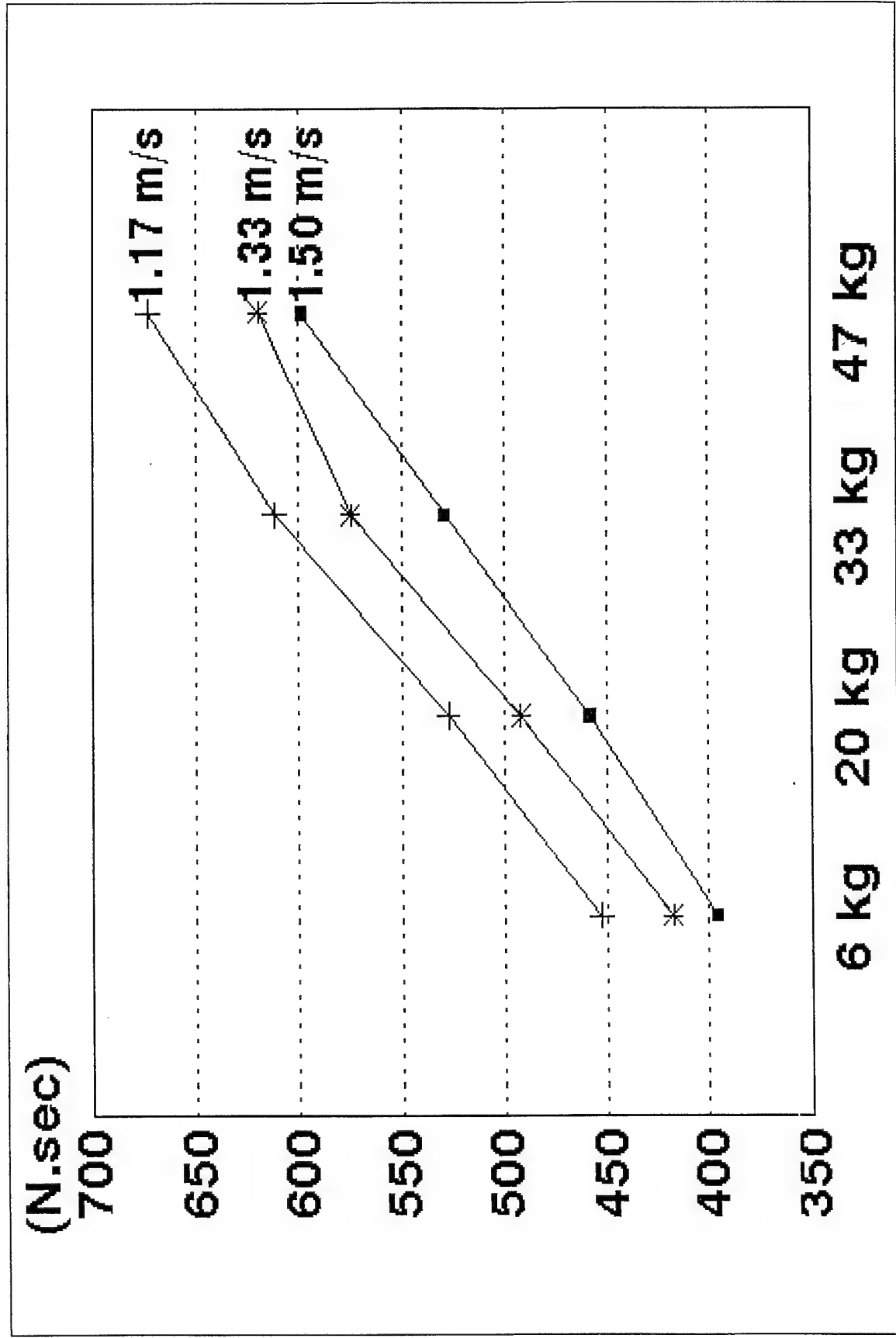


Figure 18. Interaction effects for downward thigh-on-shank impulse.

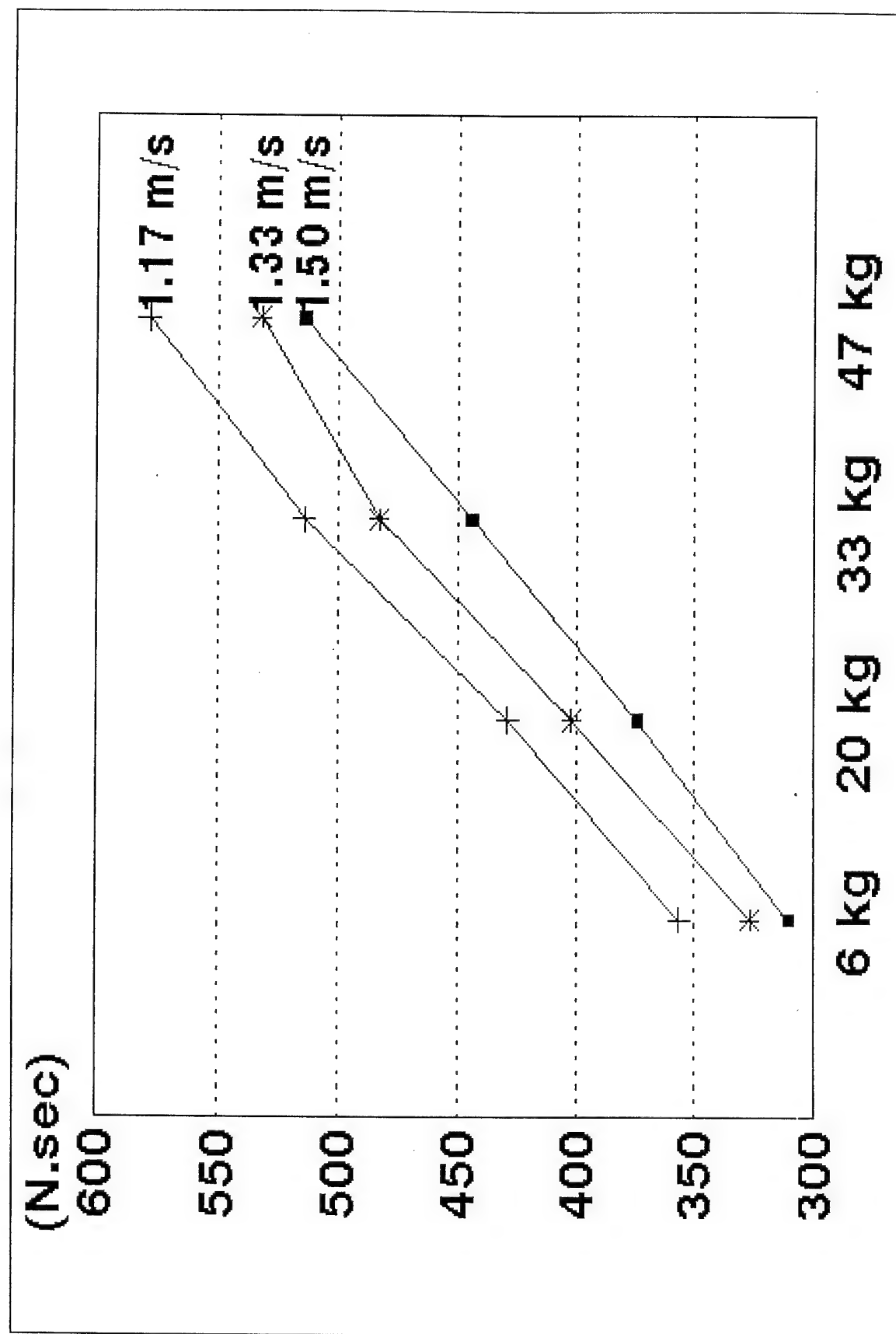


Figure 19. Interaction effects for downward trunk-on-thigh impulse.

DISCUSSION

SPEED EFFECTS

Stride Parameters

Our finding that the proportion of stride that each foot was in contact with the ground (stance phase) decreased as test speed increased was not in agreement with the data of Winter (58), which showed no difference in duration of stance phase between a slow and natural cadence while walking without load. This discrepancy may have been due to a difference in walking speeds between the two studies. In our study walking speed was fairly well controlled and consistent. In Winter's study, even though cadence was controlled by a metronome, actual walking speed was not regulated, and could even have been the same for both the slow and natural cadence, due to adjustments in stride length. Control of cadence alone is not enough to control walking speed because of the lack of control of stride length.

Lower Body Sagittal Plane Ranges of Motion

In order to compare the kinematic results of this study to those of others, joint angles were converted from radians to degrees (one radian equals 57.3°). There were no other studies found in the literature in which various walking speeds during load carriage were compared. Thus, our main comparisons were to a study by Winter (58), which biomechanically compared unloaded walking at different speeds. To translate from the ankle angle system used in this study to that used by Winter, 120° were subtracted from all ankle angles (see diagram for joint angles in the Methodology section). Our observation of slight ankle-flexion relative to the neutral position at heel-strike was similar to the findings of others (40, 58). Our observed ankle range of motion of 28° - 31° during a gait cycle was slightly higher than Winter's 24° - 29° (58), which could be attributed to differences in walking speed and the fact that Winter's volunteers were not carrying loads.

The greater knee flexion at heel-strike observed as walking speed increased probably helps to reduce shock. The mean knee angle of 178° prior to push-off observed in this study was slightly higher than the 172° noted in normal volunteers by Winter (58).

The more extended hip position at toe-off with increasing walking speed appears to be directly related to greater stride length; the hip extended to a greater degree as the walker pushed off further. The more flexed position of the hip observed at heel-strike as walking speed increased was similarly related to the longer stride length; the front leg reached further forward just as the rear leg extended further backwards as speed and stride length increased.

Upper Body Sagittal Plane Ranges of Motion

The decrease in minimum shoulder angle and greater shoulder range of motion observed as load carriage speed increased, indicated greater total arm swing resulting largely from increased swing in the rearward direction. As walking speed increases, propulsive ground reaction force on the push-off foot tends to rotate the body in a transverse plane because, from an overhead view, the force does not pass through the body's center of mass. For example, when a walker pushes off the right foot, the ground reaction force acts to rotate the body in a counter-clockwise direction when viewed from above. Newton's law states that the body's clockwise rotational momentum must thus increase. The swinging of the arms allows this momentum to increase without awkward rotation of the trunk.

Center of Mass Parameters

The longer stride length associated with greater load carriage speed accounted for the associated lower minimum center of mass vertical position. When the rear leg pushed backward as the body moved forward over a greater distance, the angle of the leg relative to the ground decreased. Thus hip height relative to the ground, proportional to the sine of that angle would have decreased as well. The greater upward and downward center of mass vertical velocities with increased load carriage speeds were necessitated by the observed greater stride frequency and vertical center of mass range of motion. With the center of mass covering a greater vertical distance in a shorter amount of time, the vertical center of mass velocity had to be greater.

Ground Reaction Forces and Impulses

The increases in most ground reaction impulses and forces were not proportional to increases in walking speed. As speed increased 28% from 1.17 to 1.50 m/s, braking impulse increased only 7%, while peak and average braking, propulsive and lateral forces increased 21%-25%, and first and second peak vertical forces increased only 6%-11%. This indicated that walking technique changed as speed increased, keeping various ground reaction forces from increasing proportionally to speed increases.

Joint Torques and Forces

It was expected that joint torques about ankle, knee, and hip would be greater when carrying loads at a faster than a slower pace. However, the degree to which the torques increased around the various lower body joints differed. The greatest percentage increase in torque occurred about the hip, and the least about the ankle, indicating that muscles producing torque about the hip were most involved in increasing load carriage speed and those producing torque about the ankle were the least involved. This was consistent with the previous finding (24) that increase in load carriage speed was effected much more through increasing horizontal force than vertical force. Given the position of the leg and foot during push-off, torque about the

ankle produces mainly vertical force, while torque about the hip produces mainly horizontal force.

Muscle Electrical Activity

For all muscle groups, the patterns of muscle activity remained the same across all walking speeds even though amplitudes tended to increase with speed. The spinal erectors showed their first peak of activity at heel-strike of the ipsilateral foot and a second peak at heel-strike of the contralateral foot. Electrical activity of both the left and right spinal erectors peaked at the heel-strikes of both the left and right feet. This must have occurred because the deceleration of the body at heel-strike was transmitted upward from the foot. The deceleration of the trunk had to be effected by the spinal erector muscles. In addition, the trunk reached its maximum forward lean just before both the left and right heel-strikes and began to rise afterwards, an action effected by the spinal erectors. It should be noted that, even though the spinal erectors on both the right and left side were active at both heel-strikes, each was most active during the heel-strike of the contralateral foot. The surface electrode placed over the right spinal erectors picked up considerably higher peak spinal erector activity at left heel-strike than at right heel-strike. This can be accounted for by the fact that ground-reaction braking force at heel-strike exerts a torque on the body, which tends to twist the trunk so that the shoulder opposite the heel-strike foot moves forward. The spinal erectors on the side opposite the heel-strike foot must contract to prevent excessive twisting of the trunk during the stride. Such work by the spinal erectors increases with the greater inertia of trunk-plus-pack associated with a heavier backpack.

Average tibialis anterior and gastrocnemius activity both increased by about 30% as walking speed increased, while quadriceps and hamstrings activity increased 40% and 80%, respectively. This greater electrical response of the flexors and extensors of the knee than those of the ankle to increases in walking speed reflects the greater increases in torque about the knee than the ankle as walking speed increases.

Backpack Acceleration

Peak upward, downward, and backward accelerations of the backpack increased with walking speed, while peak forward acceleration was not significantly affected by walking speed. A person walking at an apparently constant speed actually decelerates during and following heel-strike and accelerates during the propulsive phase. The ground reaction forces that effect these changes in speed are transmitted through the foot upward through the skeletal structure. As the trunk experiences these forces transmitted via the hips, and is accelerated or decelerated, it exerts force on the pack straps or undersurface. When the trunk accelerates in the upward, downward, or backward directions, the attachment of pack to person is firm enough so that there is not much damping of acceleration. Thus, when the trunk accelerates to a greater degree as walking speed increases, backpack acceleration in the upward, downward and backward directions increases as well.

The lack of speed effect on peak forward backpack acceleration was probably due to the fact that much of the force accelerating the pack forward was transmitted from the load carrier to the pack through the shoulder straps. Given the configuration of the straps, the trunk could move forward somewhat before the straps became taut enough to begin accelerating the pack to a major degree. The result was that forward force on the pack was damped by the straps, causing the curve of forward force application on the pack to be flatter than would be expected from the forward acceleration of the trunk, thus diminishing peak forward pack acceleration. This damping effect likely accounted for the lack of difference in peak forward pack acceleration with increasing speed. Yet because the pack was already against the back as the trunk began to move backwards, there was little or no damping of backward forces exerted by the load carrier's back against the pack. Thus there was a significant speed effect on peak backward backpack acceleration.

LOAD-SPEED INTERACTION EFFECTS

The statistically significant interaction between speed and load effects on stride frequency reflects the fact that, at the fastest walking speed, moving from the next-heaviest to the heaviest load did not result in an adaptive reduction in stride frequency, despite that fact that stride frequency did increase when progressing to the heaviest load at the two lower walking speeds. This is attributable to the difficulty of striding at a very high cadence. At the two lower walking speeds, it was not difficult for the volunteers to respond to the increase in load by reducing their stride lengths somewhat so as to increase stability and reduce joint torques, concomitantly increasing stride frequency to maintain speed. However, when volunteers were walking at the fastest speed, and stride frequency was already high, it would have been difficult to increase stride frequency further. Walking becomes very difficult at high cadence; therefore, the volunteers did not increase their cadence at the fastest walking speed even when the load became very heavy. This may have increased their vulnerability to injury due to increased leverage through which ground reaction forces acted on the bones of the skeletal lever system and decreased stability of gait.

The interactions for vertical ground reaction impulse, downward shank-on-foot impulse, downward thigh-on-shank impulse, and downward trunk-on-thigh impulse appear to be due to lower than expected impulses with the 47 kg load at the 1.33 m/s walking speed. This can be attributed to the greatest increase in stride frequency and greatest decrease in stride time at the 1.33 m/s walking speed when the load was increased from 33 kg to 47 kg. Since impulse is the area under a force vs. time curve, a shorter stride time yields a lower impulse, assuming similar forces. The lack of interaction for average or peak forces indicates that the impulse interactions were due to time effects rather than force effects.

CONCLUSIONS

Based on the data analysis, the following conclusions were drawn:

1. As load carriage speed increased there was greater knee flexion at heel-strike, which probably helped reduce shock.
2. As load carriage speed increased, the hip position at toe-off became more extended. As the rear leg pushed off to a greater degree at higher walking speeds, the front leg stretched further forward.
3. There was greater total arm swing with increasing speed, most of which was accounted for by increased arm swing in the rearward direction, which helped prevent excessive trunk rotation.
4. Lower minimum vertical position of the body center of mass occurred as load carriage speed increased, reflecting the geometry of a longer stride length.
5. The greater upward and downward center of mass vertical velocities associated with increased load carriage speeds were necessitated by greater stride frequency and vertical center of mass range of motion.
6. Load carriage technique changed as speed increased, keeping several ground reaction forces lower than proportional to the increase in speed.
7. The greatest percentage of joint torque increase with load carriage speed increase occurred about the hip, and the least occurred about the ankle, indicating that muscles producing torque about the hip were most involved in increasing load carriage speed, and those producing torque about the ankle were the least involved. The electrical activity data from the leg muscles supported the joint torque findings. In keeping with the action of the hip muscles, increase in load carriage speed was effected much more through increasing horizontal than vertical ground reaction force.
8. While the amplitudes of muscle activity tended to increase with speed, the patterns of muscle activity remained the same across all walking speeds.
9. The initial propulsive impulse seen at heel-strike resulted from flexion at the knee, rather than extension at the hip, as the heel struck the ground, and was effected by hamstring muscle activity.
10. The pack design used in the experiment, which incorporated a waist belt and frame, did not prevent the shoulders from supporting a considerable portion of the load by shifting it to the hips.

11. The spinal erectors produced their largest burst of activity at contralateral heel-strike, accounted for by the necessity for controlling the degree of trunk twisting during the stride. A smaller burst of activity at ipsilateral heel-strike was related to deceleration of the forward motion of the trunk as the body was braked at heel-strike, with the subsequent raising of the trunk.
12. The load carriage stride, in terms of the muscles that flexed and extended the knee, was characterized by concentric knee flexion from shortly before to shortly after heel-strike, eccentric knee flexion during a shock absorption phase, concentric knee extension during push-off, and a quiescent period after toe-off during the swing phase.
13. The load carriage stride, in terms of the muscles that dorsiflexed and plantarflexed the foot, was characterized by eccentric tibialis anterior activity at heel-strike which controlled the rate of plantarflexion to prevent the foot from slapping against the ground. The magnitude of tibialis anterior activity increased proportionally to speed. The gastrocnemius was largely inactive except for a period of high activity during push-off, which occurred between mid-stance and heel-strike of the contralateral foot and included the second peak for vertical ground reaction force.

RECOMMENDATIONS

LOAD CARRIAGE TECHNIQUE

1. Excessively increasing downward force on the ground when attempting to increase load carriage speed can have the undesirable effect of raising the body and load higher than is necessary, thereby increasing the work involved in load carriage. In addition, any augmentation of vertical forces at the feet is transmitted upwards through the skeletal structures including the foot, ankle, tibia, fibula, knee, femur, hip, pelvis, and lower back, with possibly injurious effect.
2. When increasing load carriage speed, it is desirable to allow the knees to flex through a greater range of motion after heel-strike in order to improve shock absorption.

PHYSICAL EXERCISES TO IMPROVE LOAD CARRIAGE PERFORMANCE

1. Because the hip extensors become increasingly important as load carriage speed increases, strengthening of the gluteus muscles through resistance exercise may help increase the speed at which a given load can be carried. Exercises that can be used to strengthen the hip extensors include the barbell or Smith machine squat, barbell or dumbbell lunge, barbell or dumbbell step-up onto a 30-46 cm (12"-18") high box, walking up stairs 1-3 steps at a time while wearing a weighted belt or vest, leg-press, and hip extension on a weight-stack machine. While less important for increasing speed, general load carriage ability can also be improved by exercising the knee extensors. Some of the exercises recommended above for the hip extensors also work the knee extensors. In addition, work on a knee-extension weight-stack machine provides specific exercise for the knee extensors.
2. Both the spinal erector and abdominal muscles should be strengthened to improve general load carriage ability. Exercises for the spinal erectors include the face-down trunk raise while holding a weight plate against the chest with the legs stabilized in either a horizontal or 45° position, the good-morning barbell trunk raise, and the barbell dead-lift. General exercises for the abdominals include sit-ups, curl-ups, and leg raises (the first two can be weight-resisted). More specific supplemental exercises can be added which closely match the body segment ranges of motion seen in load carriage. These would include relatively short range of motion (~10°) hip and trunk flexion and extension centered on a trunk position of about 7°-14° of forward trunk inclination. Limited range of motion sit-ups and face-down trunk raises with or without added weight would be beneficial. The trunk twisting observed during the stride may be simulated while performing such exercises.

3. Because the normal stride involves both concentric and eccentric muscle activity, exercise machines that have no eccentric phase are not recommended to improve load carriage performance. Such machines include those where resistance is provided by air or fluid cylinders, and isokinetic or near-isokinetic machines that provide concentric resistance only.

LOAD CARRIAGE EQUIPMENT DESIGN

1. It appears that work is still needed to ensure that, no matter what combination of military equipment is carried, including backpack, load carrying vest, body armor, weapons, ammunition, survival gear, communications devices, etc., the pack can be adjusted to comfortably distribute the load between the shoulders, back, and hips, without excessive point pressures anywhere.
2. It may be beneficial to provide additional damping for deceleration of the pack's forward velocity. This may be accomplished by placing energy absorptive material between the pack and the back.
3. Because the second peak vertical ground reaction force (at push-off) is higher than the first (shortly after heel-strike), the addition of cushioning material in the shoe forefoot may help reduce possibly injurious forces to the skeleton. The cushioning material should be elastic so that absorbed energy can be utilized in propulsion, thereby avoiding an increase in the energy cost of load carriage.

RECOMMENDATIONS FOR FUTURE STUDY

Because very few studies have been conducted on the biomechanics of load carriage, there are many possible areas of further investigation. The effects on load carriage biomechanics of the following variables should be examined:

Equipment

- pack type
- footwear type
- clothing
- load distribution
- objects carried

Individual differences

- gender
- height
- body mass
- body proportions
- age
- physical fitness
- psychological status

Physical status

- level of fatigue
- female hormone cycle
- hydration status
- body temperature

Technique

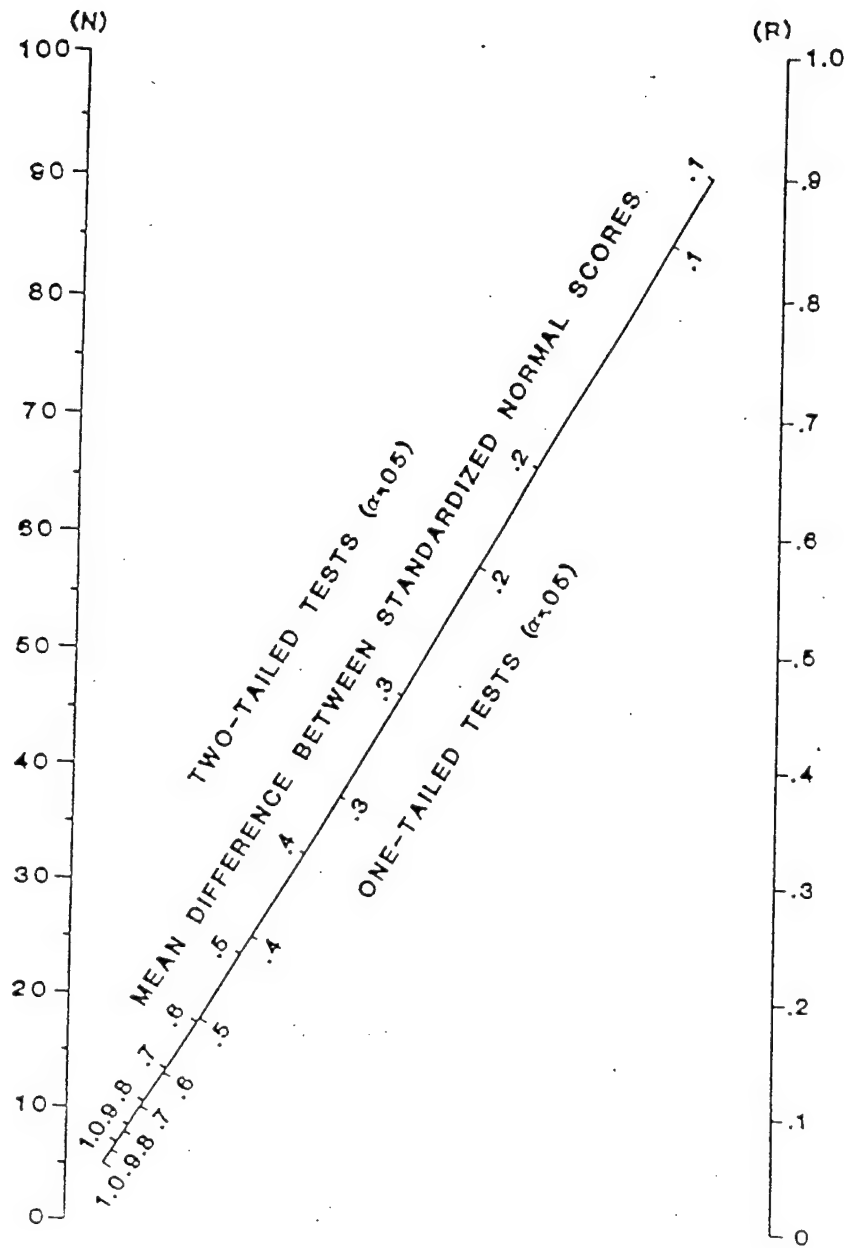
- running vs. walking
- stride length
- stride frequency

Environment

- terrain type
- surface incline
- ambient temperature
- ambient humidity
- ambient air pressure
- wind speed
- local acceleration of gravity
- altitude

Appendix A

Nomogram used for sample size estimation



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